Development of a direct mechanical ventricular assist device

by

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CHAPTER 1. INTRODUCTION

The cardiovascular system serves as a transport mechanism for the body. It consists of a four chambered heart and two closed loops-- the systemic circulation and the pulmonary circulation. The heart, essentially a pump, is responsible for supplying blood, the transport medium, to each of the circulatory loops. The right side of the heart pumps blood through the pulmonary circulation to the lungs where it receives oxygen from the inspired air and expels carbon dioxide to the expired air. The left side of the heart pumps blood through the systemic circulation to the periphery and specialized organs such as the kidney and the liver which allow for the adequate maintenance of nutrients and removal of waste products. The supply of nutrients and removal of waste products is essential to maintain the biochemical environment of the individual cells. If the cardiac output is insufficient for prolonged periods, homeostasis is disturbed and a life-threatening situation results.

The trauma of thoracic surgery occasionally results in a weakened heart which is unable to maintain an adequate workload or cardiac output. Without cardiac assistance, the condition of the patient may degrade further and the heart could suddenly or gradually fail. The two most common solutions to increase cardiac output and to decrease the workload of the heart until it sufficiently recovers are the extension of time on the cardiopulmonary bypass machine and the use of an intra-aortic balloon pump. During open-heart surgery, the heart is stopped by introducing a cold solution of potassium into the chambers of the heart. This allows the surgeon to perform the required operation without the constant movement of the heart. The

cardiopulmonary bypass machine takes over the function of the heart during the surgery. After the operation, the potassium solution is removed and the heart begins to contract again. If the force of contraction does not appear to be adequate, the surgeon might leave the patient on the cardiopulmonary bypass machine until the heart has recovered further and is contracting stronger. Another option is to use an intra-aortic balloon pump. The intra-aortic balloon catheter is introduced into the descending aorta and is inflated during the diastolic phase of the cardiac cycle. This counter-pulsation helps pull the blood out of the left ventricle during the subsequent systolic phase and thus helps increase the output of the left ventricle. However, extending the time on the cardiopulmonary bypass machine or using an intra-aortic balloon pump is successful only about 50% of the time.

Several devices called artificial left ventricular assist devices or "ALVADS" are used when the intra-aortic balloon pump offers insufficient cardiac support. The insertion of ALVADS is a very difficult surgical procedure which involves bypassing the normal pumping of blood by the ventricle into the ascending aorta. Instead, an artificial pump is connected from the ventricular apex to the descending thoracic aorta. Thus, great care must be used to prevent neurological and cardiac damage in these sections of the heart. In addition, ALVADS are difficult to maintain, and they require high levels of anticoagulants to prevent thrombosis formation on the surface of the devices.

A device is needed which can give partial or, in times of complete failure, total assistance until the heart is able to fully recover. In order to resolve the problems of other assist devices, the device should be able to be easily and quickly applied and removed; it should be applicable for short-term (a few hours) and long-term (a few days) periods of assistance; it should not require the use of anticoagulants; it should

prevent any cardiac rhythmic problems or fibrillation; it should not cause neurological injury and cardiac tissue damage; it should prevent embolization problems; and it should preserve other organ function in the post-operative period.

From a study conducted by Dr. Ronald K. Grooters, a cardiovascular surgeon at the Iowa Methodist Medical Center, it was estimated that at least 6 of 250 patients in a 6 month period might have benefited from such a device. Iowa Methodist, located in Des Moines, Iowa, has a prominent thoracic surgery division where 4 surgeons have performed 3500 open-heart surgeries in the past 12 years.

The purpose of the research described in this thesis was to design, develop, and evaluate a direct mechanical ventricular assist device. Laboratory studies were conducted for the initial design and development phase. Once a final design was achieved that satisfied the desired criteria, a series of animal studies were performed to evaluate the feasibility of the device.

CHAPTER 2. LITERATURE REVIEW

In the late 1950s and 1960s, there were several attempts made to develop a direct mechanical ventricular assist device. The main objective of the devices was to squeeze the heart by applying a force to the epicardial surface thereby acting as a cardiac massager to resuscitate a failing heart. At this time, there was a considerable amount of research to determine the preferred method of cardiac-resuscitation: closed-chest or open-chest. With open-chest cardiac-resuscitation showing consistently better hemodynamic results, several mechanical devices were developed to take over manual cardiac massage, the standard form of open-chest cardiac-resuscitation. Although most of the mechanical devices performed better than both open-chest manual cardiac massage and closed-chest resuscitation, closed-chest resuscitation remained the preferred method to use since opening the chest to gain access to the heart caused additional trauma to an already weakened patient. By 1970, most research in this area diminished.

One of the first methods attempted to mechanically massage the heart was performed by Bencini and Parola (1956) using a technique called "pneumomassage". This technique consisted of rhythmically insufflating the pericardial cavity with gas. Since the pericardium is a relatively strong and inelastic membrane, insufflation of the pericardial cavity resulted in compression of the heart, as opposed to inflation of the pericardium.

To ensure the strength of the pericardium was adequate, several initial studies were performed on canine and human cadavers to measure the maximum pressure allowable inside the pericardial cavity before the pericardium burst. The maximum

pressures obtained (680 - 1,125 mm Hg) were well above the estimated pressure required to compress the heart (< 200 mm Hg).

The apparatus used to introduce the gas into the pericardial cavity was a specially designed metallic cannula. After the cannula was inserted into the pericardial cavity through a small hole in the pericardium, a screw device was used to partially seal the hole and to hold the apparatus in place. A gas source was then connected to the cannula.

In order to evaluate the technique, experiments were performed on 20 mongrel dogs. After the thoracic cavity was entered and the heart exposed, ventricular fibrillation was induced without injuring the pericardium. The apparatus was then introduced into the pericardial cavity, and rhythmic insufflation was started. Maximal blood pressure in the femoral artery increased from 0 to 50-100 mm Hg and was maintained throughout the period of assistance (up to 60 minutes). A left cardiography showed that the rhythmic compression of the heart did not interfere with coronary blood flow. In each case, defibrillation was achieved after the period of total assistance.

Although this method provided adequate cardiac output during fibrillation, its usage is very limited. During most open-heart surgeries, the pericardium is incised and removed from the heart to expose the epicardium. Thus, this method of cardiac support would not be possible to resuscitate a heart after open-heart surgery. In addition, although the pericardium is a tough membrane, it will fatigue after extended use and would not be reliable for long periods of assistance. Since the atria are contained within the pericardium, they would be exposed to the same high pressures required to compress the ventricles. This would result in elevated atrial pressures which is undesirable.

Vineberg (1957) reported the development of a "mechanical heart massager". The device was designed to completely encapsulate the heart using a nylon pouch and to compress the ventricles using two inflatable rubber diaphragms sewn to the interior walls of the pouch. In order to position the device around several different size hearts, two straps of non-stretchable cloth were incorporated into the middle of the pouch, and a drawstring was provided at the top and bottom. The device was slipped underneath and wrapped around a heart, and then the cloth straps were tied to fit the pouch around the heart. The drawstrings were used as adjustments to enclose the top and bottom of the heart, respectively. Since the top drawstring was pulled tightly along the atrioventricular groove of the heart, it prevented the device from slipping off the heart during assistance.

A pump was built to apply the pulsating air pressure to the diaphragms. The pumping action was designed to match the cardiac cycle of one-third systole and two-thirds diastole. Each pump cycle had a one-third positive pressure stroke (inflation) followed by a two-thirds negative pressure stroke (suction). To accommodate different heart rates, the speed of the pump was variable.

The "mechanical heart massager" was tested on five medium-sized dogs by turning off the respirator after the thorax was entered and allowing anoxia to result. This was continued until two minutes after the heart completely stopped beating (usually about 10 minutes). At which time, the "mechanical heart massager" was placed around the heart and total assistance was started. The arterial blood pressure abruptly increased and was maintained above 80 mm Hg in each subject. With the assistance, each heart resumed beating and each of the animals completely recovered with no epicardial or myocardial damage present.

This device appeared to be promising, but no further research was ever reported. A possible problem with the device would be the difficulty of completely enclosing several different sized and shaped hearts with only a couple of straps and drawstrings to ensure that the inflating balloons would be directed into the ventricles instead of just filling the gaps between the pouch and the heart. A low ratio of stroke volume to balloon volume would most likely result especially since, even though nylon is non-distensible, it would still have a tendency to bulge out upon inflation of the diaphragms.

Wolcott et al. (1960) devised a simple, but effective means of artificially contracting the ventricles. Their "mechanical heart massager" was made with a rigid metal outer shell and a flexible rubber inner liner sealed to the edges of the outer shell. The heart was slipped inside the outer shell and alternating positive and negative pressure was applied to the space between the outer shell and the inner liner by a centrifugal pump. The inward expansion of the inner liner squeezes the heart and subsequent contraction of the inner liner allows the heart to relax for ventricular filling.

The device was evaluated using 9 mongrel dogs. To induce cardiac arrest, obstructive asphyxia was used in 4 of the animals, and the other 5 were given a slow intravenous injection of 15% potassium citrate. The device was slipped around the heart and held in place manually. After mechanically massaging the heart for periods ranging from 1 to 45 minutes, spontaneous cardiac action returned in all of the animals. From gross observation, the heart and lungs appeared to be well oxygenated, and the heart had no apparent epicardial or myocardial damage.

The "mechanical heart massager" was demonstrated to be an effective means of resuscitating a failing heart. Unlike the other devices previously mentioned, the "mechanical heart massager" does not completely enclose the heart. The cylindrical shape of the outer shell leaved the apex exposed. This eliminated the need to fit the irregular shape of the heart. Only the circumference of the outer shell and that of the largest cross-section of the heart would have to be matched. However, fitting the circumference of the outer shell to the heart might still require a variety of different sized outer shells. Another problem, which the authors didn't address, was the need to hold the device on the heart, other than manually, for long periods of assistance.

Kline (1962) outlined the development and evaluation of a myocardial prosthetic device in his doctoral dissertation. The device was intended to be used as a chronic fixture on a irreversibly failing heart and to act as an artificial means of cardiac support.

The device consisted of an inelastic, yet flexible outer shell with an expandable inner liner made from Dermoid and Guardex materials. The device was designed to encapsulate the entire heart by inserting the heart into the top opening of the flexible outer shell. The size and shape of the outer shell was determined through anatomical studies performed on both cadavers and live subjects. Since the heart size varies depending on the weight of an animal, only dogs weighing approximately 10 kg were used to prevent the need for several different sized outer shells to be made. A stem was incorporated into the outer shell to allow gas to enter and inflate the inner liner. Once the device was positioned around a heart, a hole was trephined in the sternum to allow access of the stem outside the thoracic cavity. It was then connected to a specially designed pump to obtain alternating positive and

negative pressures inside the inner liner. The outer shell was form-fitted to the heart, and the device was prevented from being ejected during assistance since the top of the shell follows the tapered contour of the atrioventricular groove of the heart. In addition, the device was stabilized since the stem was supported by the sternum.

The performance of the device was evaluated on one mongrel dog. A left thoracotomy was performed to expose the heart. Fibrillation was induced, and the heart was then slipped inside the outer shell. Total assistance was started and the carotid arterial pressure rose from 0 to 95 mm Hg. After 54 minutes of total assistance, the heart was defibrillated, and the subject fully recovered.

One of the problems with this device is the necessity to form-fit the device around a heart of a particular size and shape. This would be suitable if there were enough time to fabricate the device for a particular patient, but during times of emergency, a device would have to be readily available. The individual variances in heart size and shape would require an impractical amount of different sized and shaped devices. Thus, this device would only be feasible if it were used on a patient with imminent irreversible heart failure until a heart transplant could be performed.

One of the most elaborate mechanical assist devices, the biventricular cardiac assistor, was designed by Kolobow and Bowman (1965). The device consisted of a rigid Lexan outer shell, strong rubber ventricles, and a fluid liner. The outer shell was designed to fit the shape of the ventricles during diastole while the rubber ventricles were designed to fit the ventricles during systole. The rubber ventricles were adhered to the outer shell at the apex, and were hollow inside to allow for expansion and contraction. Only alternating negative pressures are applied to the inside of the rubber ventricles. During diastole a large negative pressure (500 mm

Hg) was applied to the inside of the rubber ventricles which caused it to be sucked up against the outer shell. This allowed for ventricular filling. During systole a less negative pressure (100 mm Hg) was applied to the inside of the rubber ventricles which caused it to relax and thus squeeze the heart. To account for slight variations and ensure a proper fit from heart to heart, a thin rubber liner was adhered to the inside of the rubber ventricles. The thin liner was then injected with fluid to fill any gaps between the heart and the rubber ventricles. The base of the rubber ventricles had a "jelly collar" which was designed to fit in the atrioventricular groove. A separate air line, which connected at the apex and extended through the rubber ventricles, was used to suck the ventricles into the assistor. This was intended to keep the device from being ejected from the ventricles during assistance.

The device was tested on mongrels dogs. After exposing the heart and removing the pericardium, fibrillation was induced; the device was positioned on the heart; and assistance was started. During assistance, the arterial pressure was maintained at 100/70 mm Hg with a heart rate of 110 beats per minute. (The control arterial pressure was 125/95 mm Hg with a heart rate of 150 beats per minute.) The atrial pressures were normal throughout the assistance period which strongly suggests that the assistor was able to maintain a balance between the right and left output of the ventricles. After assisting for an hour, the heart was successfully defibrillated.

Although the device was able to sustain adequate arterial pressure during assistance, examination of the heart showed epicardial abrasions caused by improper configuration of the rubber ventricles. The right ventricular outflow tract was frequently ejected from the assistor which indicates that fitting the "jelly collar" to the atrioventricular groove was difficult. Attempts to alleviate this problem by

lining the inside of the thin rubber liner with jelly-like "Sylgard" diluted with Silicone fluid were unsuccessful.

The "mechanical pulsator" was developed in 1966 by Zajtchek et al. It consisted of a plastic shell which enclosed the heart except for an oblong opening near the base which allowed for the passage of the great vessels. The plastic enclosure had a slot running from the base to the apex to allow the apparatus to be slipped into place and then securely closed. A curved rubber bellows was affixed to the inside of the plastic enclosure at the position of the left ventricle. An air line passed through the plastic enclosure and connected with the bellows. Pressure pulses were supplied to the bellows through the air line. A small rubber button fastened to the bellows allowed for fixed angular location.

The device was tested using mongrel dogs. After exposing the heart through a left thoracotomy and removing the pericardium, the device was positioned around the heart. The pump was then started, and the inflation of the bellows was synchronized with the action of the ventricle by adjusting the inflation and deflation points on the pump. At which time, fibrillation was induced and total assistance begun.

The mean pressures in the femoral artery during assistance ranged from 55 to 105 mm Hg with periods of assistance of 1.5 to 6.5 hr. After the assistance was stopped, defibrillation was successful and each of the subjects fully recovered. Post-mortem histological examination of the heart, lung, kidney, and liver showed no tissue damage.

Although the device demonstrated adequate hemodynamic results, no further research was ever reported. However, as previously mentioned, complete enclosure of the heart makes it difficult to apply the device to hearts of varying size and shape.

The most notable direct mechanical ventricular assist device, which is still being researched today, is the Anstadt cup. It was developed by George Anstadt and first reported in 1966 by Anstadt et al. Since then, several researchers have evaluated the Anstadt cup for various applications (Anstadt and Britz, 1968; Coogan et al., 1969; McCabe et al., 1983; Melvin et al. 1973; Skinner et al., 1967b). The Anstadt cup consists of a glass housing with an internal polyurethane diaphragm. The glass housing is formed to fit the shape of the heart. The pneumatically driven device applies alternating positive and negative pressure to the heart through an air line connected to the diaphragm. The device is held in place on the heart by a sustained negative pressure (vacuum) between the diaphragm and the heart through a separate air line which connects through the apex.

In one study on 11 mongrel dogs, total cardiac support was maintained during ventricular fibrillation on 3 of the subjects for periods of 2-3 days (Anstadt et al., 1971). Subsequent defibrillation was successful and each of the 3 dogs fully recovered. However, all of the dogs showed good hemodynamics throughout the assistance. Complications, such as a mechanical fatigue in the diaphragm and a pneumothorax, prevented the other dogs from surviving the post-operative period. In one case, where the dog was supported for 3 days, the arterial and venous pressures were maintained at the initial control values of approximately 100 mm Hg and 5 mm Hg, respectively. The cardiac output ranged from 52 to 83% of the control value. The subject fully recovered and was clinically sound 2 years after the operation.

Another study using the Anstadt cup was conducted to determine if prolonged mechanical assistance had any damaging effects to the epicardium or myocardium (Skinner et al., 1967a). Twelve mongrel dogs were assisted for periods of 6 to 24

hours during ventricular fibrillation. After the cardiac support period, the ventricles were defibrillated, and the animals were allowed to recover. Six of the dogs, which were assisted for 6 hours, were sacrificed after 48 hours. The other six dogs, which were assisted for 6 to 24 hours, were sacrificed after 3.5 to 20 months. Immediately following euthanasia, an autopsy was performed, and all the major organs were examined.

All abnormal pathological findings were limited to the heart. In all six dogs assisted for 6 hours and sacrificed after 48 hours, a discrete subendocardial ecchymosis was present in the anterior wall of the right ventricle. The ecchymoses, which were located 1 to 2 cm proximal to the pulmonic valve, ranged from mild hemorrhages to large hematomata measuring from 2 to 4 cm in diameter. Small foci of coagulated necrosis were found in the myocardium contiguous to the ecchymoses and in the subepicardial zone of all the chambers.

A layer of fibrous tissue was present in the epicardium of the dogs assisted for 6 to 24 hours and sacrificed after 3.5 to 20 months. However, this thickening did not appear to be constrictive. Scars were found in the anterior wall of each right ventricle. These were associated with the ecchymoses found in the six dogs sacrificed after 24 hours. Only a single microscopic focus of myocardial fibrosis was present in the left ventricle.

The predominant myocardial necrosis was present in the anterior wall of the right ventricle. This was most likely due to the compression of the ventricles during systole. Subvalvular pulmonic stenosis, related to the constriction of the right ventricle, has been observed during direct mechanical ventricular assistance. However, the damages inflicted on the heart were not sufficient to limit the applicability of mechanical assistance.

The Anstadt cup was the first device to be used for partial assistance on a "weakened" heart (Skinner, 1971). To simulate a "weakened" heart, the left circumflex coronary artery was ligated for 5 hours in 13 mongrel dogs. During the ligation, the mechanical assistance was synchronized to the beating heart. Of the 13 dogs, 12 survived 24 hours after the procedure, and 8 survived 1 week. A survival rate of only 33% was found in control animals given supportive treatment. During the synchronized assistance, both coronary sinus flow and myocardial oxygen consumption decreased significantly. This showed evidence that the myocardial work load can be reduced during mechanical assistance.

The Anstadt cup has proven to be an effective cardiac assist device. Several clinical trials on dogs have been conducted over the past 15 years to evaluate the performance of the device and to determine its effects on the heart and the other major organs with promising results. However, only limited clinical trials on humans have been performed (Baue et al., 1968). A possible reason for the lack of further development and subsequent application to human subjects is the difficulty of fitting the device to hearts of varying size and shape. In several cases, assistance was interrupted because the Anstadt cup was ejected from the heart. This was caused by an improper fitting of the device. In other cases, severe damage was inflicted on the myocardium because the device was too loose. An assist device must be capable of being applied to several different sized and shaped hearts. Otherwise, an excessive number of devices would be required to fit all possible hearts, and application of the device would be cumbersome and time-consuming. In addition, commercial production of the device would be unfeasible.

Only the Anstadt cup was tested as an assist device for supplementing cardiac output on a "weakened" heart. However, any of the devices could have been used as

a partial assist device with the proper timing circuits to synchronize the action of the device to the beating heart. From a hemodynamics point of view, all of the devices appeared to satisfactorily replace the function of the heart during cardiac insufficiency by maintaining adequate arterial pressure and cardiac output. The results of these past studies support the idea that direct mechanical ventricular assistance is feasible.

CHAPTER 3. DEVELOPMENT OF THE ASSIST DEVICE

The initial development of the direct mechanical ventricular assist device which is described in this report consisted of deciding upon a list of design criteria and reviewing available materials which could be used to fabricate the device. The second stage of development involved designing and constructing several prototypes. During this early phase, a laboratory set-up with an artificial ventricle was used to evaluate the performance of each design. After a working prototype was obtained, animal studies were conducted for further refinement until the final design was achieved.

From the literature review of problems encountered during previous attempts of direct mechanical ventricular assistance, two basic characteristics of the assist device to be developed were set before the design process was started. First, the device would not enclose the entire heart, since this would require the use of several different sized devices with possibly different shapes. Instead, it would cover the base and mid-section of the heart, and the apex would remain exposed. Second, two localized balloons centered at each ventricle would be used to apply the compressing force to the heart rather than squeezing the entire heart with one large lining or bladder. Direct mechanical assistance has been shown to cause some damage to the heart tissue (Skinner et al., 1967a). By applying a localized force, only localized damage should occur which is believed to be better than damaging the entire surface. Thus, it was envisioned that the basic design would include a

flexible but not distensible outer shell containing two balloons in contact with the right and left ventricle, respectively.

Design Criteria

Before the design and development of the assist device was started, the following design criteria were specified to ensure the final design satisfied the required characteristics.

1. The device should be applicable to hearts of varying size and shape so that only a few different models would be required for all cases.
2. The device should be capable of being quickly and easily placed into proper position on the heart.
3. The device should remain in proper position throughout the assistance period without the use of sutures or suction.
4. The device should not significantly interfere with venous return, arterial output, heart contractility, and/or coronary blood flow.
5. The device should be capable of delivering enough force to adequately pump the heart.
6. The device should cause minimal damage to the heart tissue.

Materials and Equipment

Since the assist device was constructed in a research laboratory and numerous prototypes were required to be tested, complex and costly methods of production

were not feasible, and thus, the materials chosen had to ensure that fabrication of the device was relatively simple.

Two of the main materials used during the initial development of the device were silicone rubber and Silastic medical grade sheeting. These materials were chosen because they can be readily bonded to other silicone rubbers and Silastic sheetings and reactive metals, such as aluminum, using a Silastic Type A medical adhesive. In addition, several types of silicone rubber are available with a range of distensibility. Silicone rubber reinforced with fiberglass (1/16 in. thick) and silicone rubber reinforced with a Dacron mesh (1/16 in. thick) were tested in the portions of the assist device which were required to be nondistensible, the outer shell and the balloon backing, respectively. Eventually, a sheet of 0.005 in. thick stainless steel was used for the outer shell. In the inflatable sections of the assist device, sheets of non-reinforced Silastic with thicknesses of either 0.01 or 0.02 in. were used.

A Datascope, Inc. System 83 intra-aortic balloon pump was used to apply the alternating positive and negative pressure to the balloons. The machine is normally used to inflate an intra-aortic balloon catheter for diastolic augmentation, and it is commonly used for pressurizing 70 mL of air or helium into a 40 mL balloon volume. The balloon pump system contains timing circuitry to allow the inflation and deflation of the balloon to be triggered by the QRS complex of the ECG. During standard operation, the intra-aortic balloon catheter is inflated during diastole, thus a predetermined delay of 125 ms is set between the R-wave of the ECG and the start of inflation. However, this delay can be interrupted so the inflation of the balloon can be triggered immediately following the R-wave. This feature is imperative for direct mechanical assistance to allow the inflation of the balloon to occur during systole. Separate inflation and deflation controls are

available for further adjustment and fine-tuning to match the balloon inflation to the contraction of the heart for optimal assistance. The inflation point can be adjusted from 0 to 500 ms after the R-wave, and the deflation point can be adjusted from 3 to 625 ms after the inflation point. In addition, the balloon can be triggered to inflate on every beat, every other beat, and every third beat. The balloon pump system also has an internal trigger which causes the inflation-deflation of the balloon to occur at a constant rate. The rate can be adjusted from 60 to 130 bpm. This is useful during ventricular fibrillation when a QRS complex is not present to trigger the inflation-deflation of the balloon. Another feature of the system is a balloon augmentation dial which controls the amount of gas which is pumped into the balloon. The volume of gas can be adjusted from 0 to a maximum 70 mL. This offers the capability of adjusting the level of assistance.

Although the volumes available for pumping were limited, the use of the intra-aortic balloon pump was very convenient for the initial development of the assist device. The timing circuitry makes it adaptable so that it can be used as both an intra-aortic balloon pump and a pump for a direct mechanical ventricular assist device. Without it, a pump and control unit would have to be designed and constructed. This would have been costly and time-consuming.

Laboratory Studies

During the initial design and development of the assist device, laboratory studies were conducted using an artificial ventricle to evaluate the pumping action of each prototype. This approach proved to very useful in developing the initial design since much "trial and error" was involved.

Laboratory set-up

The laboratory set-up of the artificial ventricle is shown in Figure 3.1. The artificial ventricle was made from a Dow Corning HS II RTV high strength moldmaking silicone rubber-kit and molded into an approximately conical shape, similar to an actual heart. The rigidity of the silicone rubber was increased by incorporating strips of nylon mesh laterally inside the walls of the ventricle. The top of the ventricle was closed-off using a circular piece of Plexiglas with three openings. Two of the openings were connected to 5/8 in. inlet and outlet tubes which contained one-way valves to ensure flow would only occur in the desired direction. The other end of the inlet tube was connected to the inlet reservoir which had a pressure head corresponding to the filling pressure of the ventricle. The other end of the outlet tube was connected to the outlet reservoir which had a pressure head corresponding to the workload required to pump water out of the ventricle. The overflow of the outlet reservoir was directed into the inlet reservoir to make a closed-loop system which would not require a water source. The third opening in the Plexiglass top was used to introduce a catheter inside the ventricle to measure intra-ventricular pressure.

The intra-ventricular pressure was measured using a Millar micro-tip pressure transducer (model #PC-350) along with a Millar transducer control unit (model #TCB-100). To measure the output of the ventricle during assistance, a Transonic 5/8 in. in-line ultrasonic flow probe (model #12N41) was placed in the outlet tube. The flow probe was then connected to a Transonic ultrasonic bloodflow meter (model #T101) to process the signal into a recognizable flow waveform. A continuous recording of the pressure and flow waveforms was obtained using an Astro-Med Dash II strip-chart recorder.

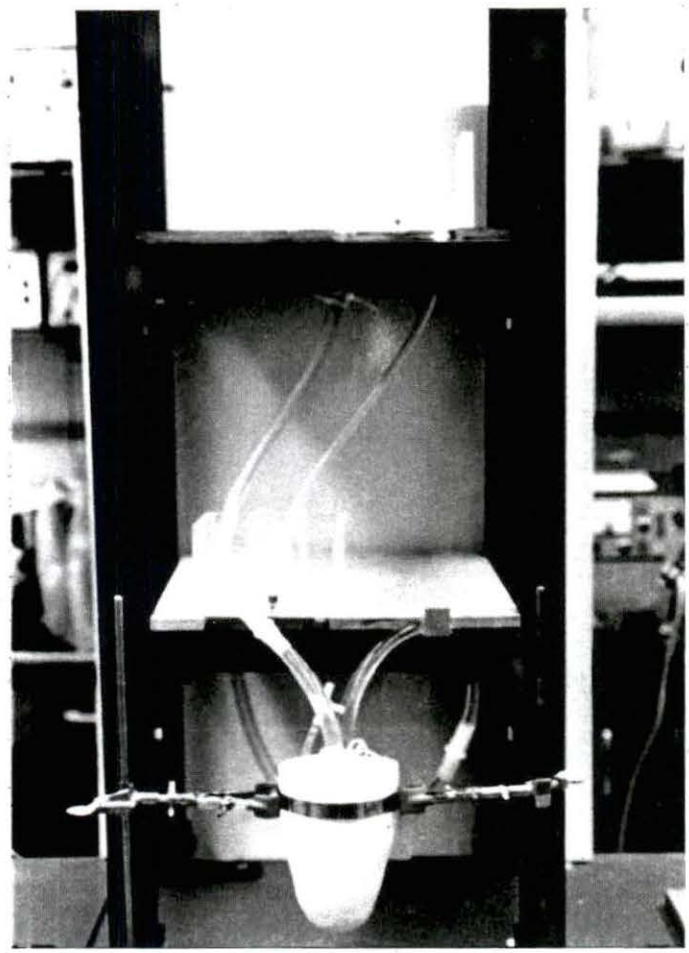


Figure 3.1. A photograph of the bench top set-up used during the laboratory studies

Design process using the artificial ventricle

The final design of the assist device evolved through many different trials. In this section, a description of this evolution during the laboratory studies is presented.

The first prototype built was a simple conical shaped cuff designed to approximately match the shape of the artificial ventricle. The outer shell was made

from a sheet of fiberglass reinforced silicone rubber. It was cut from the sheet so that it could be wrapped around the artificial ventricle, adjusted to the proper size, and then held in position by Velcro straps. The width of the outer shell was 7 cm which covered approximately half of the surface area of the ventricle. An additional Velcro strap was strung over the top of the artificial ventricle and connected to the outer shell on each opposite end. Without the strap over the top, the device would slowly slip off the ventricle during assistance. Two circular patches (5 cm dia.) of 0.01 in. Silastic sheeting, with connecting airlines, were glued to the inner surface of the shell for the balloons. Air was supplied to the balloons through tubes which passed through the outer shell and connected to the balloon pump.

This prototype successfully compressed the ventricle, and a steady flow of water through the outflow tube was observed. A recording of the intra-ventricular pressure and flow in the outlet tube is shown in Figure 3.2. However, this design was intended only to demonstrate the use of the laboratory set-up. The actual shape of a real heart is irregular, and a simple conical shaped cuff would not fit a heart properly to give efficient pumping action.

The next design consisted of two circular balloons (5 cm dia.) with backings attached to a thin strap. The strap and the balloon backings, which were strengthened with a strip of metal, were made from Dacron reinforced silicone rubber. Dacron reinforced silicone rubber was used instead of the fiberglass reinforced because it was less bulky. Since an actual heart is irregularly shaped, it was thought that a thin strap would be easier to fit around the ventricles rather than a shell which wrapped around the entire base and mid-section. However, when the device was placed on the artificial ventricle and assistance was started, minimal flow was observed through the outflow tube. The expansion of the balloons caused the

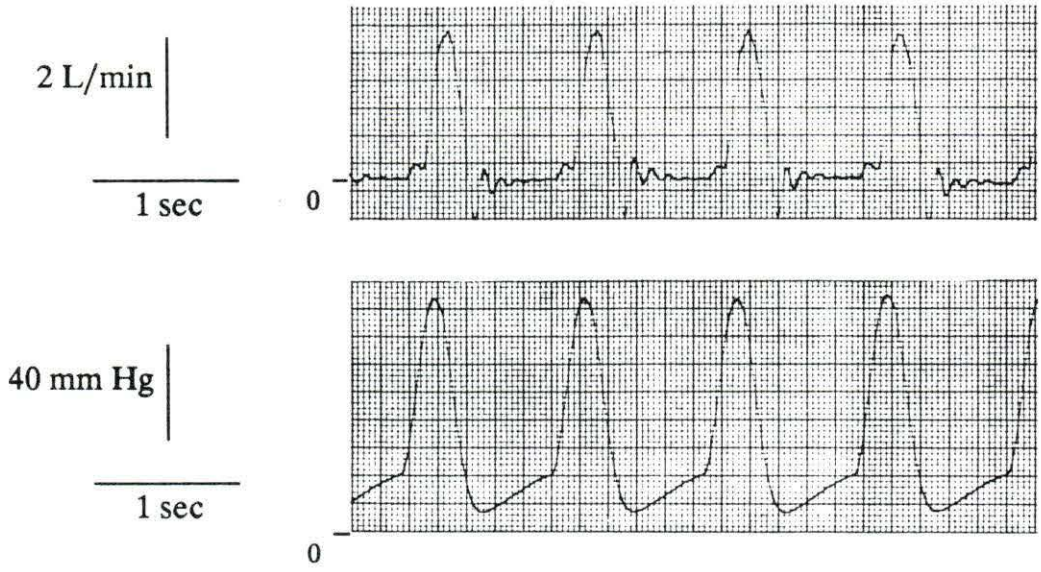


Figure 3.2. Outlet flow (top) and intra-ventricular pressure (bottom) waveforms for the artificial ventricle during assistance by the fiberglass reinforced silicone rubber conical cuff

backings to rotate around the strap. Thus, the backings did not supply enough lateral support to direct the balloons inward, and the ventricle was not being compressed.

To offer more support to the backings, a second prototype was built which had two thin straps. Instead of just one strap at the center, a strap was connected along the backings at the top and bottom. This ensured that the top and bottom of the backings would remain against the ventricle during the expansion of the balloons. When placed on the ventricle, the inflation of the balloons successfully compressed the ventricles, and a pulsatile flow and pressure signal was observed. However, the inflation of the balloons caused the middle of the backings to bulge out. Thus, only

a portion of the air supply to inflate the balloons was being used to compress the ventricle.

A third prototype was then constructed which had three thin straps: one at the top, center, and bottom of the backings. The three straps along with the strip of metal for additional support prevented the backings from bulging outward during inflation of the balloons, but this prototype was bulky and cumbersome. The additional straps made it very similar to the original conical cuff made from fiberglass reinforced silicone rubber.

Keeping in mind that a cuff with a small width would be easier to fit on an actual heart than a cuff which covered the entire base and mid-section, another conical cuff was built using the Dacron reinforced silicone rubber with a width of only 4 cm (compared to 7 cm previously). Two long rectangular patches of 0.01 in. silicone rubber were adhered to the cuff for the balloons. These "longitudinal" balloons were made so the volume upon inflation would be approximately the same as the 5 cm diameter circular balloons used previously. This ensured a proper comparison of pumping action between different prototypes. When the cuff was placed on the artificial ventricle and the balloons were rhythmically inflated and deflated, the compression of the ventricle resulted in a peak intra-ventricular pressure and outlet flow comparable to the conical cuff made from fiberglass reinforced silicone rubber with circular balloons. However, as before, the center of the cuff bulged outward upon inflation of the balloons.

It was clear that the Dacron reinforced silicone rubber was not rigid enough to offer efficient pumping action when the balloon volume was compared to the stroke volume of the ventricle during compression. To demonstrate this, a second conical cuff made from fiberglass reinforced silicone rubber was placed around the Dacron

reinforced silicone rubber cuff during assistance. A photograph of this prototype is shown in Figure 3.3. The additional support from the fiberglass reinforced silicone rubber cuff prevented bulging and larger pressures and flows were observed. A comparison between the flow and pressure waveforms with and without the fiberglass reinforced silicone rubber "support" cuff is shown in Figure 3.4 and 3.5.

The next several cuffs were made using the fiberglass reinforced silicone rubber for the outer shell in order to maintain the necessary support required to keep the full inflation of the balloons directed into the ventricle. As mentioned previously, a fiberglass reinforced conical shaped cuff could not adequately fit an actual heart,

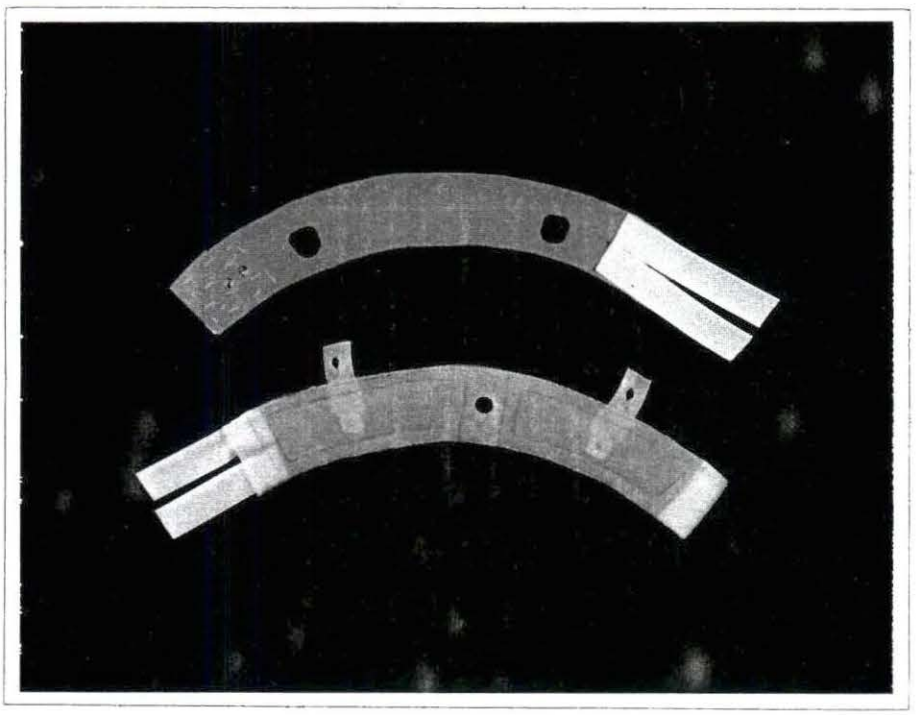


Figure 3.3. A photograph of the Dacron reinforced silicone rubber "balloon" cuff and the fiberglass reinforced silicone rubber "support" cuff

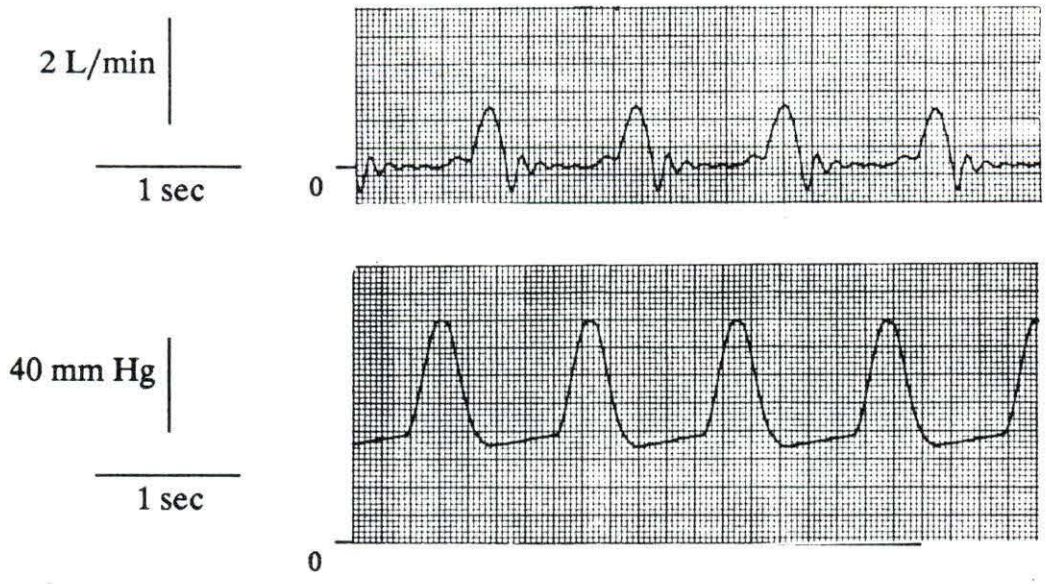


Figure 3.4. A recording of outlet flow (top) and intra-ventricular pressure (bottom) without the "support" cuff

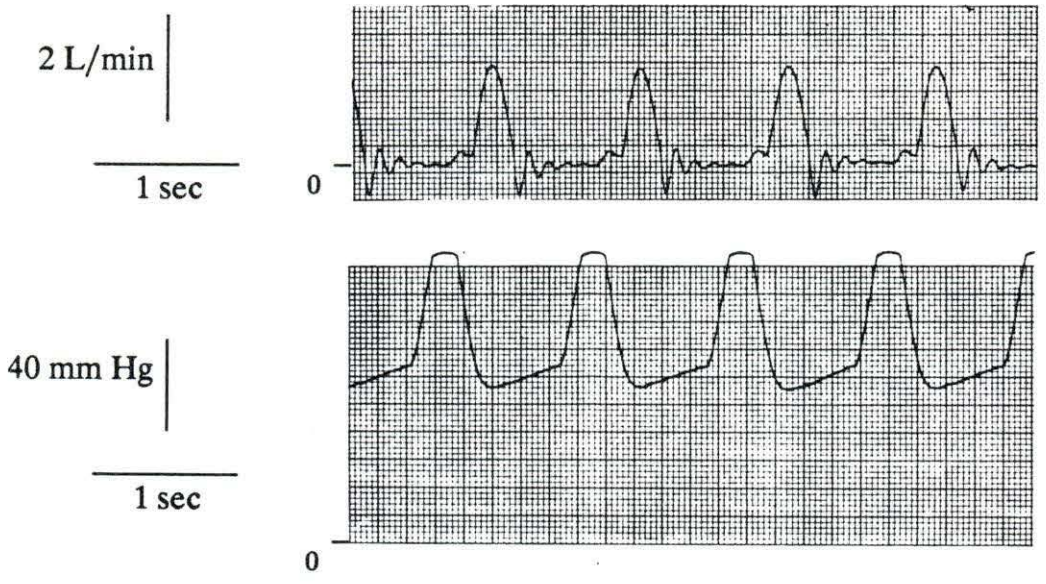


Figure 3.5. A recording of outlet flow (top) and intra-ventricular pressure (bottom) with the "support" cuff

and thus, the balloons would not be completely against the ventricles. Upon inflation, part of the balloon volume would be wasted to fill the space between the conical shaped cuff and the irregularly shaped heart. In an attempt to compensate for the poor fit, a bladder was added to the inner surface of the outer shell to fill any resulting gaps. Once the cuff was in position around a heart, the bladder could be inflated with saline until the balloons were adjacent to the ventricles. The expansion of the bladder was adjustable depending on how much saline was injected so the bladder could fit different shaped hearts. The bladder was made by glueing a sheet of 0.02 in. Silastic along the edges of the fiberglass reinforced silicone rubber shell. A valve was incorporated into the bladder for the injection and removal of the saline.

Since the bladder would make it possible to help fit the cuff to several different shaped hearts, the idea of using long rectangular balloons to decrease the width of the cuff was abandoned, and "localized" balloons were used on subsequent models. Although this would require the width of the cuff to increase back to the original value of 7 cm in order to keep the same balloon volume, the "longitudinal" balloons would have compressed the entire circumference of the heart. One of the main requirements of the device was to have two small "localized" balloons so any possible tissue damage would be limited to a small area. The "longitudinal" balloons might have caused damage around the entire circumference of the heart.

Using the idea of dividing the cuff into two parts, a "balloon" cuff and a "support" cuff, the "localized" balloons were kept separate from the outer shell and the bladder. The balloons were made by glueing a sheet of the 0.01 in. Silastic, the inflatable material, along the edges of a piece of the Dacron reinforced silicone rubber, the backing material. Even though the Dacron reinforced silicone rubber

bulged out previously, it was thought that the pressure inside the bladder should keep the majority of the expansion directed into the heart. Since part of the pressure to inflate the balloons is lost stretching the balloon material, it is imperative that the balloons be as large as possible to prevent unnecessary loss of balloon volume. Thus, a new balloon design was used in which the balloon backings were made into a trapezoidal shape as opposed to a circular shape since the area covered by a circle would be smaller than a trapezoid. A trapezoid was used because the heart is approximately conical, and the circumference of a cone decreases towards the point. Thus, the width of the balloon at the base should be larger than the width at the apex so it the matches the shape of the heart. Two slits were cut into the top of the balloon backing so they could be slipped onto a thin strap of Dacron reinforced silicone rubber. This allowed the balloons to be moveable. Ideally, the thin strap with the balloons would be placed around a heart; the balloons would be adjusted until they were at the center of each ventricle, respectively; the outer shell would be placed over the balloons; and the bladder would then be inflated until the balloons were adjacent to the ventricles.

This cuff design was placed around the artificial ventricle and assistance was started. Since the outer shell was tailor-made to fit the artificial ventricle, the expansion of the bladder was not necessary. In fact, when the bladder was pressurized, a drop in pressure and flow was observed. This is logical since the bladder would only constrict the ventricle and prevent proper filling.

After a few minutes of assistance, the cuff had slipped off the ventricle since no strap was placed over the top of the ventricle to keep it in position. Since a strap cannot be placed over the base of an actual heart without occluding the greater vessels, a different method is needed to prevent the device from moving during

assistance. The device slips off the artificial ventricle because of the balloon action. The force which compresses the ventricle has a upward component. Thus, the resisting force pushes the cuff downward. In an attempt to counter this downward force, the orientation of the trapezoidal balloons was changed. Instead of placing the longer side of the balloon at the base of the ventricle, it was switched to the apex. This was done because the end of the balloon with the longer side expands to a larger degree than the shorter side. Upon inflation, the balloon would act as a wedge between the outer shell and the artificial ventricle and tend to overcome the downward force. When the cuff with the new balloon orientation was put on the artificial ventricle, the inflation of the balloons still caused the cuff to slip off after a few minutes. Thus, the wedge created from the balloon expansion was not great enough, and the balloon orientation was changed back.

The next attempt to keep the device on the ventricle was to make the outer shell into a cylindrical shape. When inflated, the bladder would form a wedge between the cylindrical shaped outer shell and the conical shaped ventricle in an attempt to minimize the effect of the downward force caused by the balloons. Once in position, the large expansion of the bladder could still push the balloons against the ventricle for optimum assistance. Since the bladder would have to expand larger at the apex than the base of the ventricle to fill the gap between the conical shaped ventricle and the cylindrical shaped outer shell, the bottom of the bladder was folded under when it was glued to the edge of the outer shell while the top of the bladder was still glued flat. A photograph of the cylindrical shaped cuff is presented in Figure 3.6.

When this design was tested, the cylindrical cuff remained in position on the artificial ventricle for over 30 minutes without any additional support. A recording of pressure and flow is shown in Figure 3.7. The pressure and flow were comparable

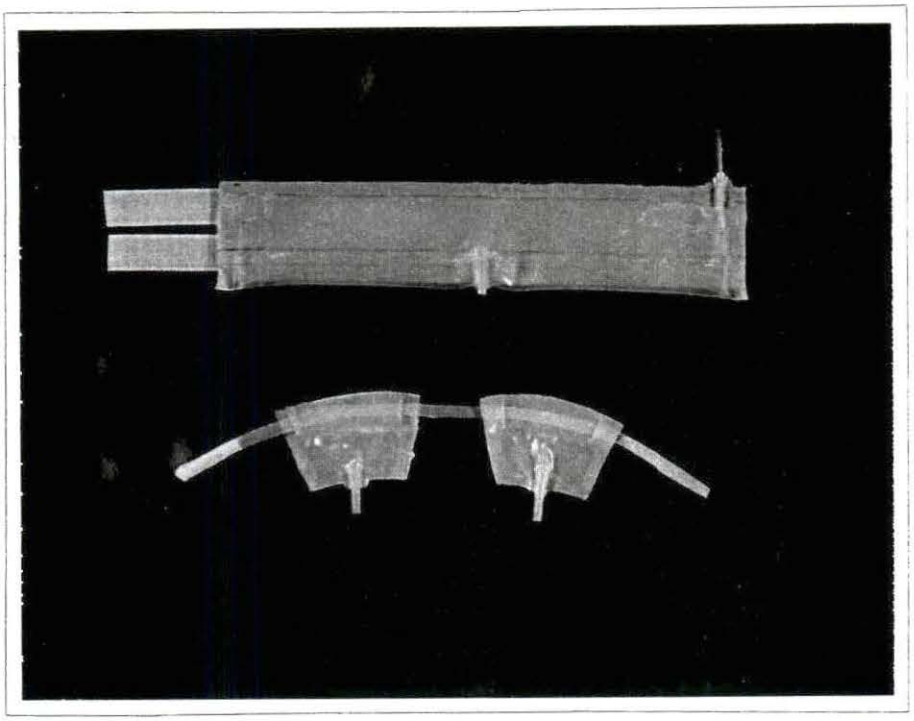


Figure 3.6. A photograph of the fiberglass reinforced silicone rubber cylindrical shaped cuff and the Dacron reinforced silicone rubber balloon strap

to that of the conical shaped cuff. Since the expansion of the bladder was adjustable, the cylindrical design should be applicable to an actual heart. Although a cylindrical shaped cuff with a bladder would still eventually slip off an actual heart, it appears to minimize the effect of the downward force caused by the balloons.

With the increased amount of saline in the bladder, the expansion of the balloons could cause displacement of the saline into the remaining portions of the bladder and allow the balloons to bulge outward. A high pressure inside the bladder would prevent the majority of the saline from being displaced, but the extra pressure inside the bladder would cause constriction of the ventricle. It was thought that if a more

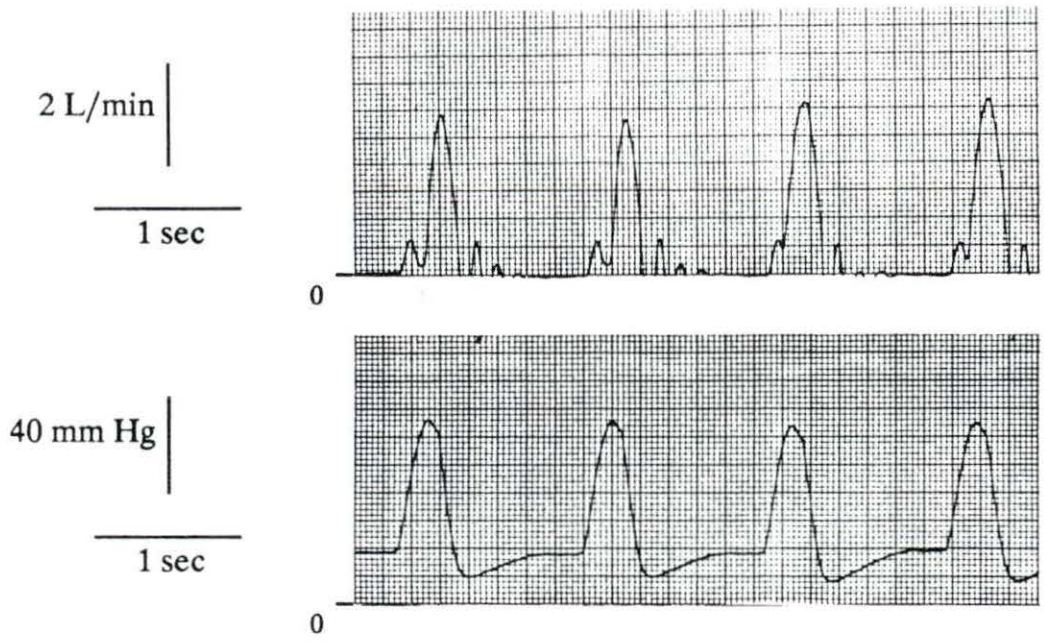


Figure 3.7. A recording of outlet flow (top) and intra-ventricular pressure (bottom) during assistance on the artificial ventricle by the cylindrical shaped cuff with bladder

viscous fluid was used, the expansion of the balloons might cause less fluid displacement. Pure glycerin was added to the bladder instead of saline. When the intra-ventricular pressure and the outlet flow were compared to the results obtained using saline inside the bladder, there was no appreciable difference. Thus, saline was used for the expansion of the bladder in the following designs because it is readily available, and if the bladder developed a leak during assistance on a patient, it should not present a problem.

The conical shaped cuff and the cylindrical shaped cuff both demonstrated adequate pumping action on the artificial ventricle. In addition, both appeared to be capable of fitting several different shaped hearts. Thus, the laboratory evaluation of the designs was promising, and the remainder of the design and

development of the assist device was directed toward animal studies involving assistance on an actual heart.

Animal Studies

During this phase of the design and development of the assist device, animal studies were performed on mongrel dogs weighing between 17 and 28 kg. The objectives of the animal studies were to ensure proper fit of the device on an actual heart; to develop a method of keeping the device in proper position during assistance; and to further evaluate the pumping action of the device. Since these criteria could not be adequately tested using a laboratory set-up, animal studies were necessary.

The prototypes were initially evaluated on hearts during ventricular fibrillation by setting the balloon pump on internal trigger so the balloons would inflate and deflate at a steady rate. These tests were used to evaluate the effectiveness of each prototype by observing the amount of assistance possible without any heart activity. Once the objectives were partially satisfied and a working prototype was achieved, the device was further tested as a partial assist device on a normal heart by synchronizing the inflation of the balloons to the contraction of the heart using the ECG trigger and timing circuitry in the balloon pump system.

Surgical procedure

The dogs were initially anesthetized with an intravenous injection of Surital (0.7 mL/kg). Positive pressure ventilation with a mixture of oxygen and halothane (2%) was used to maintain general anesthesia. An intravenous catheter was introduced

into the saphenous vein to administer fluids (lactated Ringer's solution) throughout the operation.

Originally, a median sternotomy was performed to gain access to the thoracic cavity. Following the second animal study, a left thoracotomy in the fifth intercostal space was used because it gave better exposure of the ascending aorta for placement of the flow probe. After the thoracic cavity was entered, the heart was relieved from the pericardial sac by incising the pericardium ventrally.

A surgical cut-down was performed on the left inner thigh to expose the femoral artery. A Millar micro-tip pressure transducer was then introduced into the femoral artery, and the tip was positioned in the descending aorta to measure arterial pressure. Cardiac output minus coronary flow was measured by isolating the ascending aorta and placing a Transonic ultrasonic 16 mm flow probe (model #16S572) around it. In some of the early trials, the ascending aorta was too small for the 16 mm flow probe, and a good signal was not transmitted to the bloodflow meter. Eventually, a 12 mm flow probe (model #12SB95) was purchased to be placed on the thoracic aorta. However, with this location of the transducer only a portion of the total cardiac output was measured. The transducer control unit (model #TCB-100) and the ultrasonic bloodflow meter (model #T101) were connected to the Astro-Med Dash II strip-chart recorder for a continuous recording of both aortic pressure and cardiac output. After all the surgical cut-downs were performed and before placement of the pressure transducer, heparin (50 units/kg) was injected intravenously to prevent clots from forming on the tip of the pressure transducer.

Once the flow probe and pressure transducer were in place, a control reading of aortic pressure and cardiac output was obtained. The assist device with the bladder

empty was then placed around the heart and adjusted for proper fit. Another reading was then taken to determine the effect of the device on pressure and flow. The bladder was then inflated until the balloons were adequately pushed against the heart. Another recording was taken. Ventricular fibrillation was induced by applying a low voltage alternating current directly to the myocardium. Immediately following fibrillation, assistance was started using the internal trigger on the balloon pump system. The rate of inflation ranged from 50 to 100 bpm with a total inflation time of 300 ms. A continuous recording was taken throughout the assistance period which varied depending on the performance of the prototype.

Since the artificial ventricle used during the laboratory studies was larger than an actual dog heart, a new conical cuff and cylindrical cuff had to be fabricated for the animal studies. The dimensions for the new cuffs were obtained using a plasticized heart from a dog weighing 25 kg. The conical cuff was designed by taking two circumference readings from the heart. The first reading was taken just below the atrioventricular groove. The second reading was taken 4.5 cm distal to the first reading. With these three measurements the outer shell was made to fit the approximated conical shape of the plasticized dog heart. The cylindrical cuff was made by using the measurement from the atrioventricular groove as the length of the outer shell and 4.5 cm as the width of the outer shell. The width of the outer shell for both the conical cuff and cylindrical cuff was chosen as 4.5 cm because the device covered approximately half of the plasticized heart at this dimension. This was the same protocol used during the laboratory studies with the artificial ventricle. The balloons were scaled down in a similar manner to match the smaller outer shell size. Since the heart size varies with weight, the weight of the animals was kept as close to 25 kg as possible. Therefore, these outer shell dimensions were used during

the majority of the cuff development while the cuff design changed from experiment to experiment. This prevented the need for several different sized and shaped cuffs to be fabricated for each new prototype.

Design process for the outer shell and balloons using total assistance

The first few trials were used to experiment with the fit of the balloon strap and the outer shell and to determine the pumping action of the conical shaped cuff. In one trial when the balloon strap and the outer shell were positioned around the heart and the bladder was inflated, the aortic pressure dropped from 65/45 mm Hg to 40/30 mm Hg. (The 16 mm flow probe was not working properly at this time because it was too large to fit the ascending aorta so no measurement of aortic flow was available.) Readjustment of the balloon strap and the outer shell to increase pressure had no effect. However, pressurization of the bladder had a drastic effect on the pressure depending on the degree of inflation. The drop in pressure was believed to be caused by the extra weight of the cuff on the heart and the constriction of the ventricles by inflation of the bladder. The recordings showing aortic pressure before and after the device was placed on the heart are shown in Figures 3.8 and 3.9

When ventricular fibrillation was induced and total assistance was started, the aortic pressure increased from 0 to 40/32 mm Hg-- the pressure before fibrillation. A recording of aortic pressure during total assistance is shown in Figure 3.10. These pressures were maintained throughout the assistance period of 15 minutes. During this time, the cuff was held in place on the heart manually. Although these pressures were not adequate for proper perfusion of the vital organs, the pumping action of this rather crude prototype was promising.

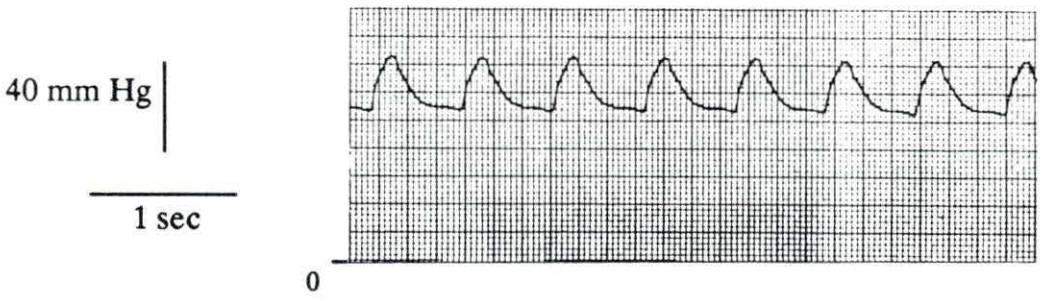


Figure 3.8 A recording of aortic pressure before the conical shaped cuff was placed around the heart (control)

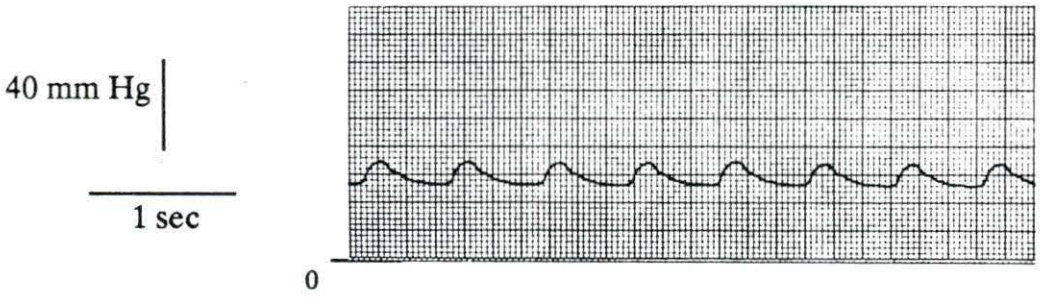


Figure 3.9 A recording of aortic pressure after the conical shaped cuff was positioned on the heart and the bladder was inflated

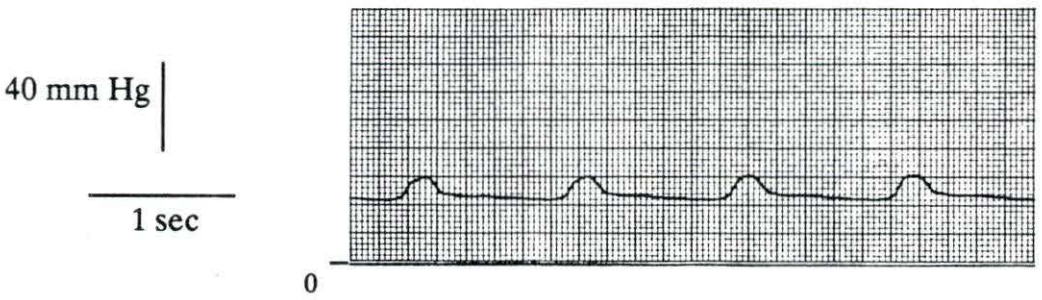


Figure 3.10. A recording of aortic pressure during total assistance by the conical shaped cuff

Investigation of the balloons during assistance showed that each balloon was not expanding to the expected volume of 20 mL upon inflation. (The balloon pump is capable of compressing 70 mL of air into a total balloon volume of 40 mL.) Since the size of the balloons had to be decreased to match the smaller sized cuff, the possible expansion of the balloons was limited. The balloon pump could not supply enough pressure to stretch the patch of 0.01 in. Silastic sheeting to a volume of 20 mL. Instead, the air supplied was being compressed into a smaller volume.

Hand massage was performed after the assistance period to demonstrate that larger mean pressures and pulse pressures are possible when the heart is squeezed more forcibly. The aortic pressure was increased to 75/30 mm Hg which was larger than the control value.

In order to compress the heart more forcibly, the balloons needed to expand to a larger volume. Originally, the balloons were made by glueing a sheet of 0.01 in. Silastic flat along the edges of a piece of Dacron reinforced silicone rubber. To increase the expansion of the balloons, the sheet of 0.01 in. Silastic was folded under before it was glued to the top and bottom edge of the balloon backing. The right and left sides were still glued flat. In addition, the width of the cuff was increased from 4.5 to 5.5 cm which allowed the balloon size to increase accordingly. With this increase in width, the cuff covered about two-thirds of the surface area of the heart. These modifications greatly increased the expansion of each balloon from approximately 10 to 17 mL.

The next couple of trials were used to compare the performance of the conical shaped cuff and the cylindrical shaped cuff. A recording of the aortic pressure and aortic flow during ventricular fibrillation using the conical cuff is shown in Figure 3.11. A similar recording for the cylindrical shaped cuff is shown in Figure 3.12.

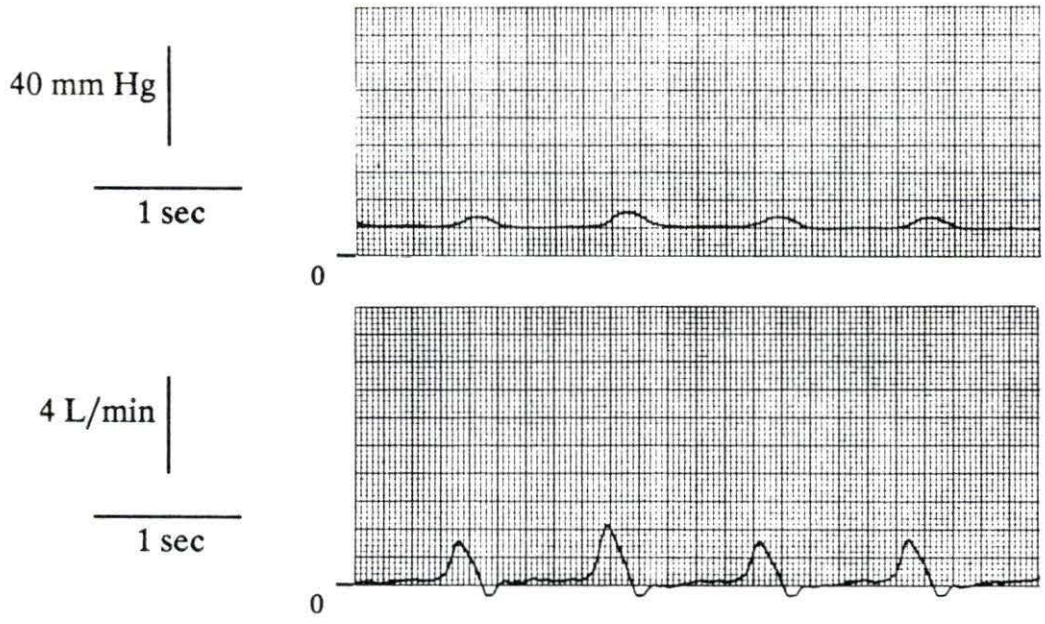


Figure 3.11. A recording of aortic pressure (top) and flow (bottom) during total assistance on a fibrillating heart using the conical cuff

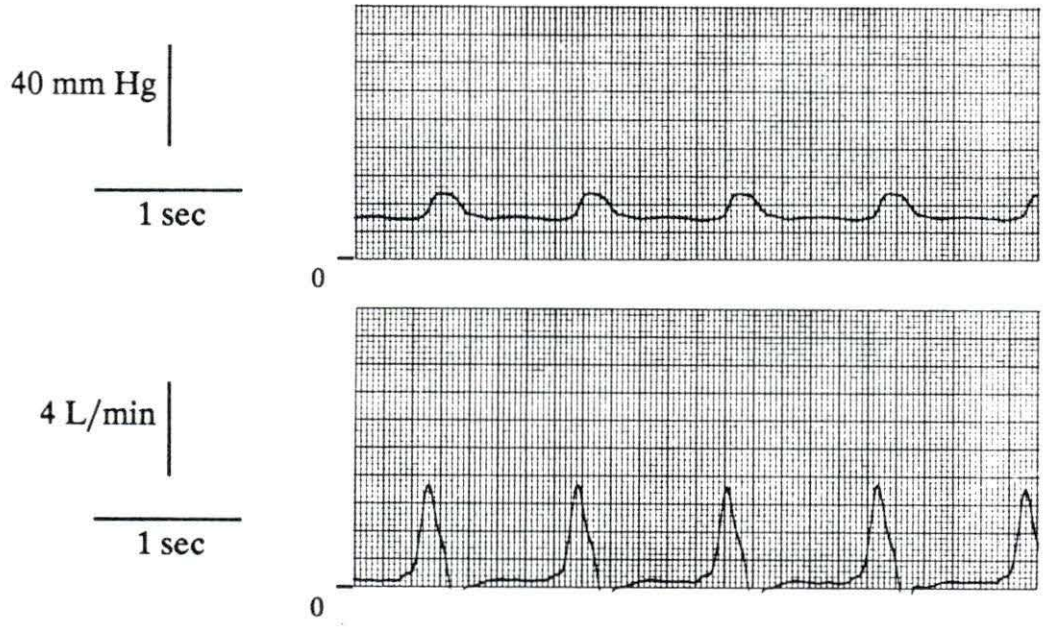


Figure 3.12. A recording of aortic pressure (top) and flow (bottom) during total assistance on a fibrillating heart using the cylindrical cuff

The aortic pressure for the conical shaped cuff was 40/30 mm Hg while the aortic pressure for the cylindrical cuff was 60/40 mm Hg. The original control reading was 85/65 with a pulse pressure of 20 mm Hg. Although the mean pressure for the cylindrical cuff was lower than the control value during total assistance, the pulse pressure was equal. The mean aortic pressure was even lower with the conical cuff and the pulse pressure was only 10 mm Hg. Thus, the pumping action of the cylindrical cuff was significantly better than the conical cuff.

The difference in performance between the two prototypes was related to the fit. The cylindrical cuff was able to fit the shape of the heart better than the conical cuff. Although the size of the hearts used in this series of experiments remained relatively constant, the shape of each heart varied. The conical shaped cuff was designed to approximately fit the plasticized heart by taking two circumference measurements which dictated the size and shape of the outer shell. If one or both of the two circumferences on an actual heart was different than the plasticized heart, the conical cuff did not fit properly, and the shape of the cuff became distorted. Inflation of the bladder only caused further distortion. On the other hand, the cylindrical cuff was designed using only the circumference from the atrioventricular groove on the plasticized heart. If an actual heart had a different circumference than the plasticized heart, the circumference of the outer shell could be easily adjusted without distortion by overlapping the ends of the cuff until the correct circumference was obtained. Velcro straps on each end of the outer shell were then used to maintain this circumference. The bladder can then still be used as a wedge to fill the space between the outer shell and the heart. Since the cylindrical cuff was more readily adjustable to obtain a proper fit on several different sized and shaped hearts, it was used in the remaining prototypes.

During the trials comparing the conical cuff and the cylindrical cuff, it was very difficult to keep the balloon strap in the same position as the outer shell. The balloon strap continually slid around between the heart and the outer shell. Eventually, the balloons moved and they were no longer centered on each ventricle. To prevent movement of the balloons, the balloon strap was fastened to the outer shell. The balloon strap was not connected to the outer shell near the balloons so the balloons could still be adjusted until they were at the center of each ventricle. This partially solved the problem, but the balloons were still able to move around since the balloon strap was not connected to the outer shell near the balloons.

Finally, to solve this problem, the balloons were fastened directly to the outer shell, and the balloon strap was removed. Although this prevented adjustment of the balloons, another method could be developed, if necessary, so that the balloons could slide along the top of the outer shell for proper positioning. However, the balloons only needed minor adjustment once the balloon strap was placed on a heart, and thus, the need to position the balloons may not be necessary as long as the balloons are close to the center of the each ventricle. The deletion of the balloon strap removed some of the bulk and weight from the device. The bulk and weight of the device was one of the reasons the aortic pressure dropped when the device was placed around a heart.

Design process for attachment of the device to the heart using total assistance

Instead of manually holding the cuff in place during the previous trials, while a method was still being developed to keep it from sliding off, the pericardium was pulled over the top of the cuff and clamped at two opposite sides of the outer shell. Since the pericardium is a tough membrane, it was able to hold the cuff in position

for periods of up to 30 minutes. However, the pericardium is subject to fatigue, and therefore, it would not be a reliable means of keeping the cuff on the heart during long periods of assistance. One of the design criteria for the device was the ability to remain in position on a heart without the use of the pericardium. Thus, this method was not considered a permanent means of cuff support.

One of the first attempts at holding the cuff in place on the heart consisted of a strap made from Dacron reinforced silicone rubber with five strips of umbilical tape connected to it at equal intervals. The strap was intended to be wrapped around the atrioventricular groove into a ring shape. Since the heart starts to taper at the atrioventricular groove, the ring would be prevented from sliding downward off the heart. After the ring was in position, the five strips of umbilical tape were clamped at the respective locations to the cuff.

Although this method was successful in keeping the cuff in position during most of the assistance period, the aortic pressure was considerably lowered when this method was implemented. The drop in pressure was most likely caused by the downward force on the ring in the atrioventricular groove which prevented the cuff from being ejected from the heart. This force probably caused a partial occlusion of one or both of the low pressure vessels terminating at the base of the heart, the vena cava and the pulmonary vein. This would inhibit ventricular filling and lower the ventricular pressure. In addition, the attachment of the umbilical tape at the anterior of the cuff was very awkward. Since this method interfered with normal heart function, it was considered to be an unsuccessful attempt of keeping the cuff in place on a heart.

A second attempt to keep the device from being ejected from a heart consisted of adhering an inflatable strip of 0.01 in. Silastic along the top of the cylindrical outer

shell. After the cuff was wrapped around a heart, the strip was inflated with air. The strip formed a tear-drop shape which extended over the base of the heart. Since the base of the heart tapers off at the atrioventricular groove, the expanded strip would be supported by the base of the heart, and the cuff would be prevented from slipping downward.

This method was unsuccessful. The inflated strip prevented the cuff from being completely ejected from the heart, but the cuff still moved downward. After 10 minutes, the cuff was out of position and the balloons were no longer centered on each ventricle. However, in previous experiments, the base of the heart had a tendency to bulge outward upon inflation of the balloons, especially at the pulmonary conus. With the expansion of the strip, the base of the heart was contained in the device and no bulging was observed. This was verified in the pressure and flow recordings. The cuff with the inflatable strip produced a larger aortic pressure and flow than the cuff without the inflatable strip; the aortic pressure was increased from 22/14 to 30/16 mm Hg while the peak aortic flow was increased from 2 to 3 L/min. Even though this method was unable to keep the cuff in proper position, it did show the advantage of containing the base of the heart to increase total compression.

Keeping in mind the importance of containing the base of the heart inside of the assist device to prevent bulging of the pulmonary conus, a new method was developed to keep the cuff in position during assistance. A very thin strip of nylon material with a piece of umbilical tape interwoven through it at the top was fastened to the top of the cylindrical outer shell. Once the cuff was placed in position on the heart, the ends of the umbilical tape were pulled tight and clamped. This nylon purse-string "dome" wrapped around the base of the heart. Since the ring of

umbilical tape was pulled tight over the base of the heart where it tapers, the cuff would be prevented from moving downward. In addition, since the nylon "dome" contained the base of the heart, the pulmonary conus was constrained.

When this method was tested, the nylon purse-string "dome" kept the cuff in position during the entire assistance period of 30 minutes with no indication that the cuff was moving. When the umbilical tape was pulled tight, there was no change in aortic pressure or flow. Therefore, it was not occluding any of the low pressure vessels terminating at the base of the heart. In addition, the nylon "dome" prevented the pulmonary conus from bulging out. This was verified through observation. To further evaluate this method, it was tested during the remainder of the animal trials.

Design process for the bladder using total assistance

Throughout the animal studies, the prototypes had a significant effect on the aortic pressure and flow when they were placed on the heart and the bladder was inflated. In some cases, the aortic pressure dropped to 50% of the control value while the aortic flow dropped by as much as 60% of the control value. In order for the assist device to be used on a "weakened" heart, the drop in pressure and flow had to be minimized. Otherwise, the device would have a damaging effect, and the condition of the heart would further deteriorate.

It was thought that the drop in pressure and flow was caused by the addition of extra weight and bulkiness to the heart and the constriction of the ventricles upon inflation of the bladder. The pressure inside the bladder has to be large enough to prevent saline displacement during inflation of the balloons, but when the bladder was being inflated, it was being expanded into the heart causing constriction. This would decrease ventricular filling, and by Starling's law, the aortic pressure and flow

would decrease. To ensure that the full inflation of the balloons is directed into the ventricles, the final bladder design should keep the balloons adjacent to the heart. In addition, to offer stability and help prevent movement, the bladder should fill the gap between the cylindrical shaped outer shell and the irregular shaped heart. To satisfy these requirements, the pressurized bladder should fit the shape of a relaxed heart.

The bladder used previously was fabricated by simply adhering a strip of the 0.02 in. silicone rubber along the edges of the outer shell. The bottom of the bladder was folded under before it was adhered. This allowed the bottom portion to expand further and fill the gap between the apex and the cylindrical outer shell. During the pressurization of this bladder, the outer shell had a tendency to change from its cylindrical shape into a triangular shape. When the outer shell is wrapped around a heart, creases form in the empty bladder. Upon inflation, the expansion of the bladder removes these creases. Since the outer shell was constrained at the ends, all the creases were removed except for three which caused the outer shell to become distorted from its intended circular shape into a triangular shape. Since a cross-section of the heart is approximately circular, the ventricles would be compressed during inflation of the bladder.

To prevent the outer shell from distorting, a hollow aluminum cylinder with a wall thickness of 1 mm was bonded to the outside of the outer shell. Since the aluminum was rigid, it remained in a cylindrical shape when the bladder was pressurized. In order for the cuff to have an adjustable circumference, a section of the cylinder was removed near the ends of the cuff. Otherwise, the new design would be limited, and it would not be able to fit several different sized hearts.

When the new prototype was placed around a heart, the pressure and flow dropped by 30% and 40%, respectively. Subsequent inflation of the bladder caused the pressure and flow to drop even further. The extra weight and bulk of the aluminum had a severe effect on the pumping ability of the heart. In addition, even though the outer shell maintained its cylindrical shape, the expansion of the bladder was still constricting the heart. Thus, the aluminum cylinder only made the original problem even worse.

To obtain a better fit of the bladder to the heart, the expansion of the bladder was controlled. This was accomplished by glueing the sheet of 0.02 in. Silastic onto the outer shell to form seven equally spaced partitioned bladder sections. The partitions between each bladder section extended downward about 4 cm from the top of the outer shell (approximately two-thirds of the total width of the cuff). The bottom portion of the bladder was not constrained by the partitions so it could still be able to fill the large gap between the outer shell and the apex of the heart. By making seven equally spaced bladder sections, the outer shell was forced into a hexagonal shape when the bladder was inflated, as opposed to a triangular shape when the bladder was unconstrained. A hexagon was considered to be a close approximation of the desired circular shape. In addition, with the partitioned sections, the bladder could be inflated to a larger pressure without constricting the heart. Thus, the expansion of the balloons would be directed into the ventricles instead of simply displacing the fluid in the bladder.

When the cuff was placed on the heart and the bladder with the new design was inflated, the aortic pressure and flow fell 35% and 40%, respectively, of the control value. This was an improvement from previous trials. When ventricular fibrillation was induced and total assistance was started at 70 bpm, the pressure increased from

0 to 40/20 mm Hg and the flow increased from 0 to 0.8 L/min. The control values were 110/60 mm Hg and 3.5 L/min. These values for total assistance were encouraging since the control values were so high, and each balloon was only expanding to a volume of 17 mL. With a balloon volume of 17 mL and a heart rate of 70 bpm, the largest cardiac output possible, if the entire balloon volume was compressing the heart, would be 1.19 L/min. Thus, with an assisted value of 0.8 L/min the balloons were pumping the heart with approximately 67% efficiency. If a larger balloon volume were available, the pressure and flow would most likely have increased in proportion to the increase in the balloon volume. The inflation of the new bladder design only slightly constricted the heart, and it seemed to keep the balloons against the heart during inflation. Thus, constraining the bladder appeared to be an effective means of fitting the cuff to the heart.

The purse-string nylon "dome" was used during the previous animal trials, and it effectively kept each prototype in proper position on the heart for periods exceeding 1 hour. Thus, this method was considered to be reliable for short-term applications. However, another method was developed in case this method would fail during extended periods of assistance. If two methods were developed that were tested and considered reliable for at least short periods of assistance (< 2 hours), the cuff would be less likely to be ejected from a heart if both were used during long-term applications. This second method of attachment is discussed in conjunction with assistance on a beating heart.

With a working prototype and a promising method to keep the device in position, the remainder of the design and development of the assist device was focused towards partial assistance on a normal beating heart.

Design process using partial assistance

During this phase, the bladder design and the outer shell design were refined, and another method was developed to keep the device in position during assistance on a beating heart (partial assistance).

In order to evaluate the remaining prototypes using partial assistance, the inflation of the balloons had to be synchronized with the contraction of the heart. This was possible by using the intra-aortic balloon pump system. The timing circuitry in the system uses the QRS complex of the ECG to triggered the inflation of the balloons. The QRS complex is an electrical representation of the depolarization of the myocardium. Immediately following depolarization, the myocardium contracts. The timing circuitry in the balloon pump system enables the inflation of the balloons to be delayed a specified time after the peak of the QRS complex is detected. In addition, the deflation of the balloons can be delayed a specified time after the point of inflation. With this timing control, the action of the balloons can be adjusted until the optimum assistance is achieved.

Since a normal ECG contains a P-wave, the QRS complex, and a T-wave, the balloon pump must only be triggered when the R-wave is detected. This is possible because, under normal circumstances, the amplitude of the R-wave is larger than the amplitudes of the P-wave and T-wave. However, if the amplitude of the R-wave is too small, or the amplitude of the P-wave and/or the T-wave is too large, the balloon inflation would be improperly triggered. Thus, a reliable ECG is imperative for proper partial assistance.

Originally, the ECG was obtained using a standard lead II configuration. However, with the thoracic cavity opened, the amplitude of the ECG was very small, and the balloon pump system was unable to detect the R-wave. Thus, the balloons

were not triggered to inflate. The closed-chest acts as a volume conductor and the ECG signal propagates throughout the body. When the chest was opened, the volume conductor was lost, and the amplitude of the signal at the appendages decreased.

To increase the amplitude of the ECG, the ECG electrodes were placed directly on the epicardium. This increased the amplitude of the ECG, but the signal was not reliable. The ECG was noisy, and the electrodes had to be constantly moved around to maintain an adequate signal. During assistance, several irregular beats were observed on the pressure and flow recordings because the balloons were being triggered to inflate at the improper time.

Finally, a good reliable ECG was achieved by introducing a catheter, with four ECG electrodes at the tip, into the left jugular vein and placing the tip near the right atrium. (This technique is normally used to acquire an ECG for a pacemaker.) Two of the electrodes were then connected to the lead I configuration on the balloon pump system. Since blood is a conductive medium, the ECG propagated through the blood, and it was picked-up by the electrodes on the tip. However, the signal received was not always a classical ECG. Since the tip was placed adjacent to the right atrium, the electrodes occasionally picked-up specific conductive pathways of the heart. Since the signal received still triggered the balloon pump and the delay between the peak detection and the inflation point was adjustable, this was not a problem.

The prototype with the partitioned bladder sections successfully forced the outer shell into a cylindrical shape when the bladder was pressurized. However, the shape of an actual heart is approximately oval. The bladder was modified to form an oval shape by changing the width of each of the seven individual sections. A longer

bladder section was used across the part of the outer shell which was positioned at the right side of the heart. Two shorter bladder sections were used to wrap the outer shell around the caudal and cranial side of the heart. This was also the location of each of the balloons. Unlike the right side of the heart, the two ends of the outer shell are adjusted and fastened at the left side of the heart. Therefore, one long bladder section could not be used to cover the left side of the heart, and two shorter bladder sections were used on each end of the outer shell. When the bladder was pressurized, the outer shell was forced more into an oval shape which would fit the heart better.

The bladder was also modified by removing the unnecessary portions of the bladder. The bladder was needed to keep the balloons against the ventricles to fill the gap between the outer shell and the apex of the heart. Thus, the top portions of the bladder sections at the anterior and posterior of the heart were removed because they were not required to support the cuff or the balloons.

In addition, the bladder sections over the balloons were tapered slightly at the top. With the top of these bladder sections tapered, the bladder would expand less. The circumference of the cuff is adjusted to match the circumference of the top of the heart, and a large expansion of the bladder to fit the heart was not needed. The largest expansion was needed at the bottom of the cuff since the heart tapers toward the apex. By removing the unnecessary portions of the bladder, the expansion of the bladder would cause less constriction of the heart.

When this prototype was placed on a normal beating heart, the pressure and flow decreased 20% and 22%, respectively, below the control value. This was a continued improvement from previous prototypes. While assisting every other beat with the inflation of the balloons synchronized with the contraction of the heart, the

aortic pressure and peak aortic flow increased from 53/46 mm Hg and 2.1 L/min with no assistance to 69/43 mm Hg and 4.9 L/min with assistance, respectively. Thus, the new bladder design was keeping the balloons adjacent to the ventricles. During assistance, the mean pressure and peak flow increased by 13% and 133%, respectively. A recording of the pressure and flow showing the control values, the drop in pressure caused by placing the cuff on the heart, and assisting every other beat is shown in Figures 3.13 - 3.15. Assisting every other beat demonstrated that the cuff could effectively be used on a beating heart to supplement cardiac output.

To further evaluate the performance of the purse-string nylon "dome", it was tested during assistance on a beating heart. During the previous trials, the nylon purse-string "dome" successfully kept the cuff on the heart throughout the assistance periods without any indication of movement. The nylon or the umbilical tape did not appear to be constricting any of the blood vessels at the base of the heart. When the purse-string was pulled tight, no drop in pressure or flow was observed.

To further decrease the bulkiness and weight of the cuff, the material used for the outer shell was changed. Instead of using the fiberglass reinforced silicone rubber, a sheet of 0.005 in. thick stainless steel shim stock was used. The use of the stainless steel cuff helped decrease the drop in pressure and flow when the cuff was placed on a heart. Although the stainless steel was 12.5 times thinner than the fiberglass reinforced silicone rubber, it offered at least the same amount of support; the stainless steel did not bulge out during the inflation of the balloons. Since the stainless steel was so thin, it was still flexible and could be easily wrapped around a heart.

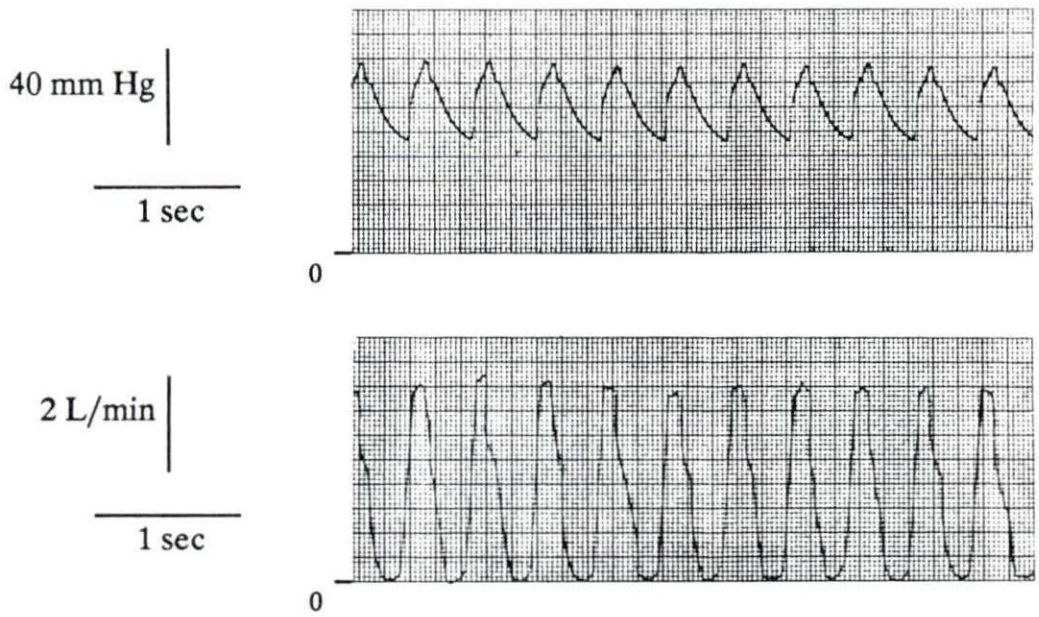


Figure 3.13. A recording of aortic pressure (top) and flow (bottom) before the cuff with the partitioned bladder was placed on the heart

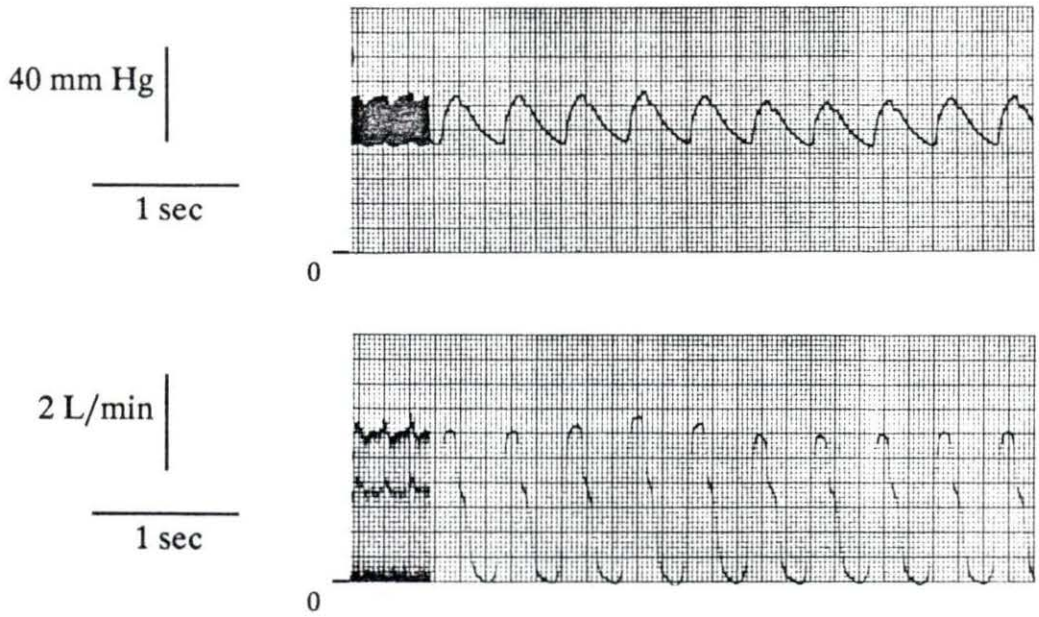


Figure 3.14. A recording of aortic pressure (top) and flow (bottom) after the cuff with the partitioned bladder was placed on the heart

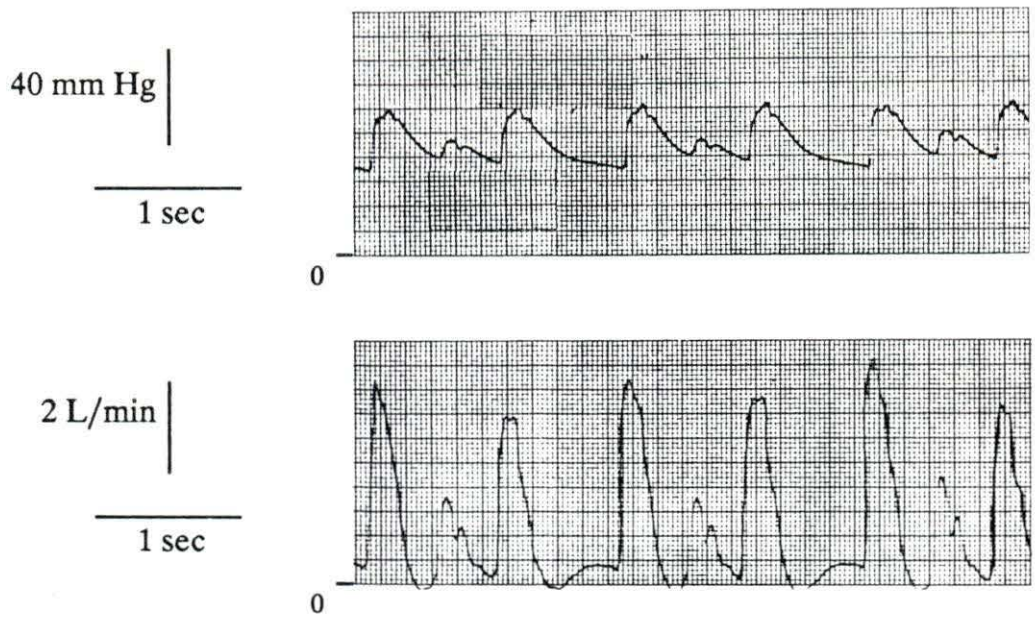


Figure 3.15. A recording of aortic pressure (top) and flow (bottom) during partial assistance (every other beat) on a normal beating heart using the cuff with the partitioned bladder

A second method to keep the device in position during assistance was developed using the ascending aorta as a fixation point. Since the aorta is a high-pressure artery and thus, fairly rigid, it should be able to support the cuff and prevent it from slipping off the heart. An "aortic ring" was placed around the ascending aorta for attachment to the device. Two small stubs of brass with a strip of umbilical tape at the end were fastened to opposite sides of the ring. Each small stub was used to support a hollow metal rod which was formed into an L-shape. Each strip of umbilical tape was slipped through a hollow L-shaped rod; the hollow L-shaped rod was then positioned on the stub; and the free end of the umbilical tape with the other end of the L-shaped rod was pulled tight and fastened to the cuff. The L-shaped metal rods were used to keep the umbilical tape off the base of the heart to

prevent occlusion of any blood vessels. The strips of umbilical tape were connected to the cuff at the right cranial side and the left caudal side of the heart, respectively. At these positions, the strips with the supports were easily accessible for adjustment and connection to the aortic ring. A photograph of the aortic ring with the straps and supports is shown in Figure 3.16.

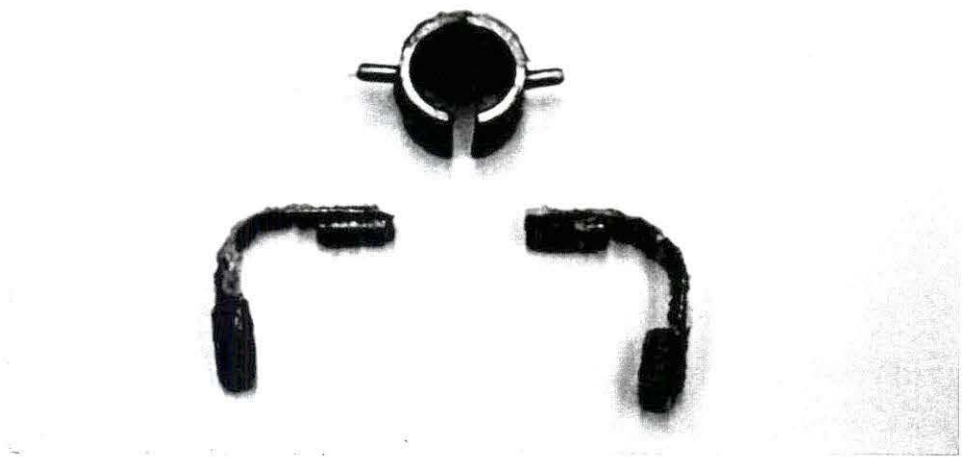


Figure 3.16. A photograph showing the aortic ring with the straps and supports.

This method kept the cuff in position on the heart for over 30 minutes during three animal trials. The force on the aortic ring did not cause occlusion of the aorta, and the supports maintained their position and prevented the strips of umbilical tape from occluding any of the blood vessels at the base of the heart. When compared to the purse-string nylon "dome", which was tested on the same animals,

there was no difference between the drop in pressure and flow when the cuff was placed on the heart or during partial assistance. This method proved to be reliable during the relatively short-term tests, and it was anticipated that it would also be reliable for long-term assistance.

With a cuff demonstrating good partial assistance and two reliable methods for keeping the cuff in position on a heart, the design and development of the assist device was completed, and the final design was evaluated using a series of carefully controlled animal studies.

Final Design

Since the device developed is designed to be wrapped around a heart, like a cuff, and assist a heart to supplement cardiac output, the term heart cuff is used to refer to the final design.

The outer shell of the heart cuff consists of a 0.005 in. thick rectangular sheet of stainless steel with a bladder made from a 0.02 in. thick Silastic sheeting. The bladder is adhered to the outer shell in partitioned sections so when the bladder is inflated, it stabilizes the cuff and keeps the balloons against the epicardial surface of the ventricles. An illustration of the heart cuff is shown in Figure 3.17, and photographs of the heart cuff are shown in Figures 3.18 - 3.20.

The two balloons are made with reinforced silicone rubber used as the backing material and 0.01 in. thick Silastic sheeting used as the inflatable material. The nondistensible backing is used to ensure that the inflation of the balloons is directed into the ventricles and not into the outer shell.

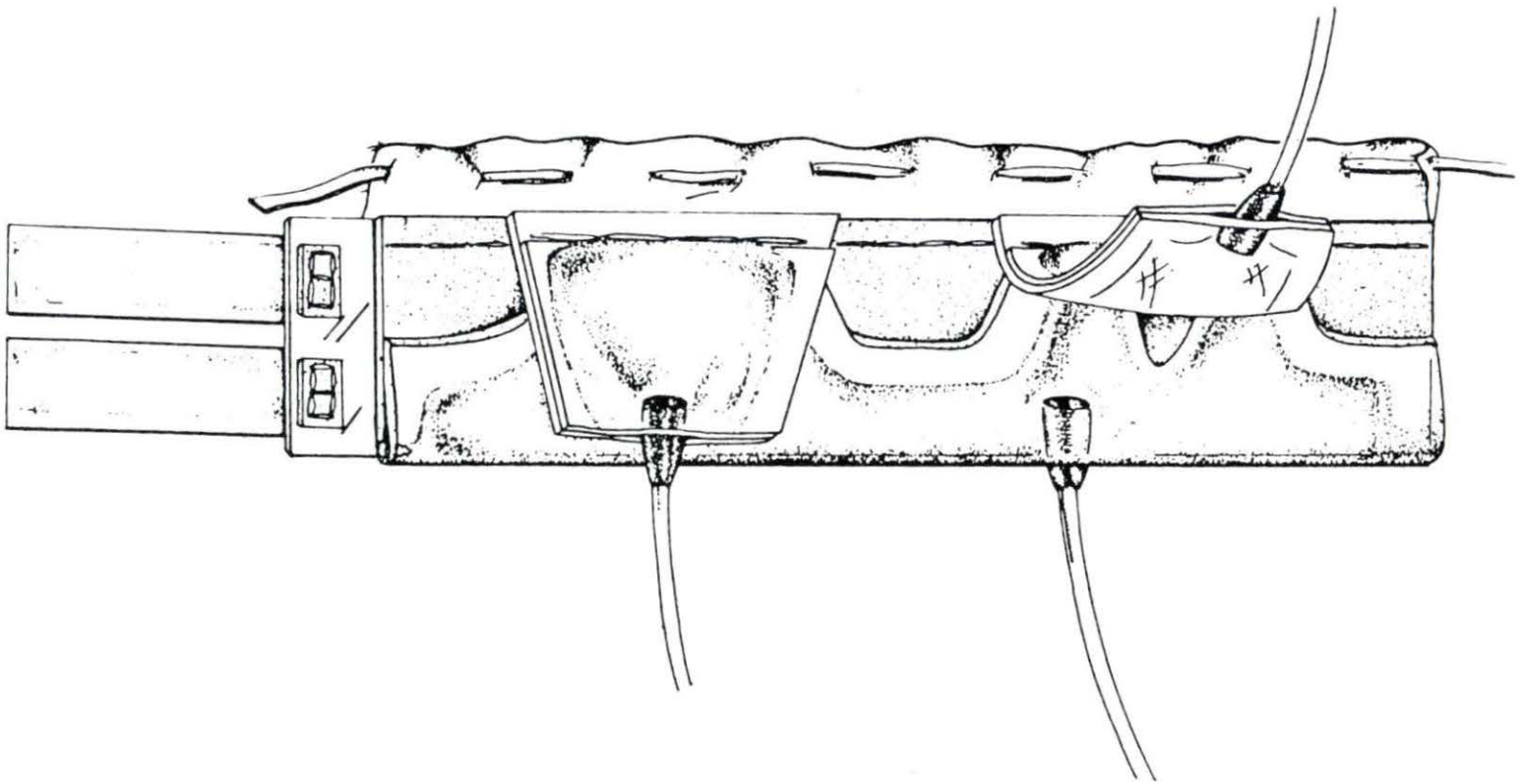


Figure 3.17. An illustration of the heart cuff with purse-string nylon "dome"

Figure 3.18. A photograph of the inner surface of the heart cuff

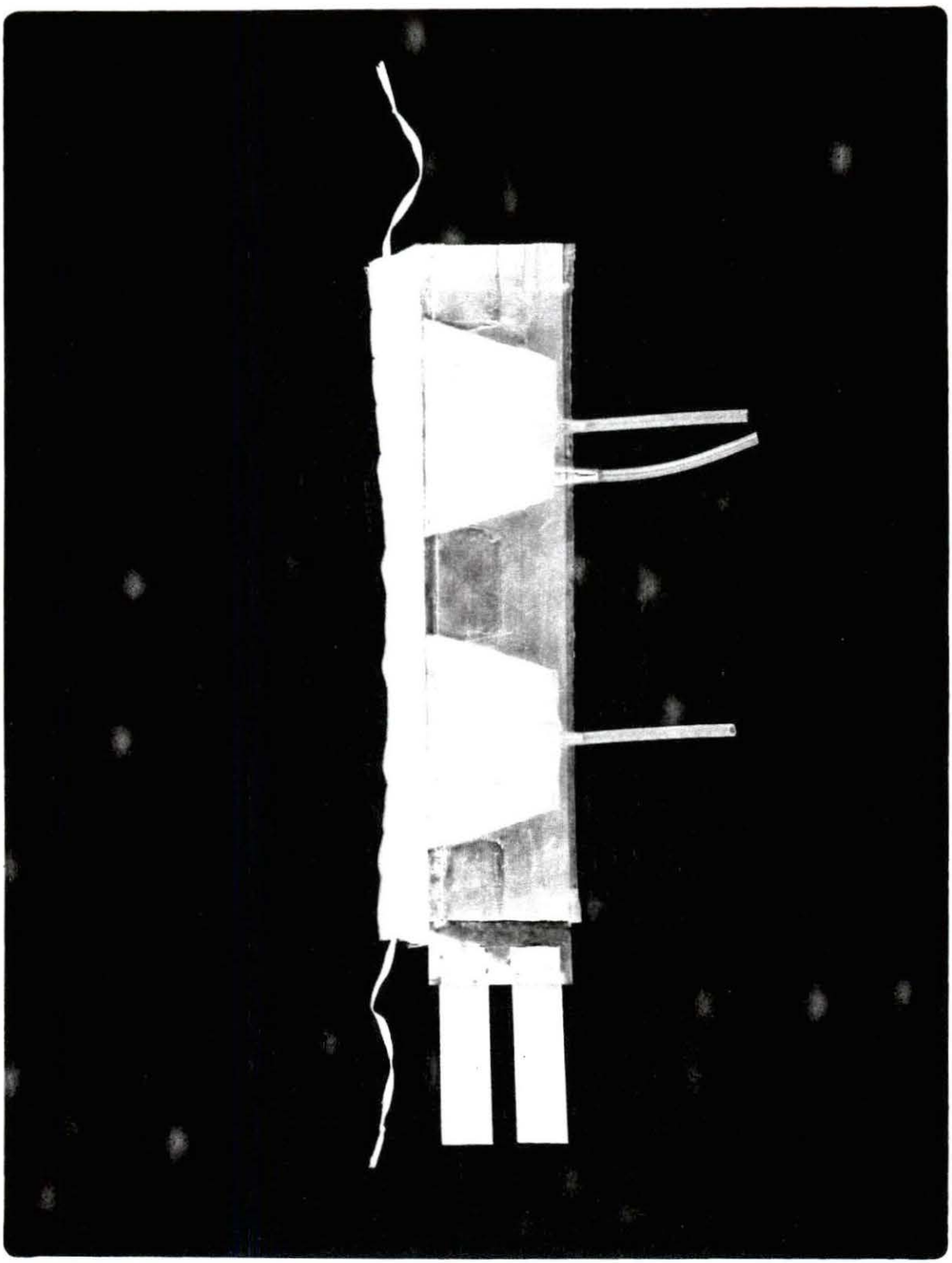


Figure 3.19. A photograph of the outer surface of the heart cuff.

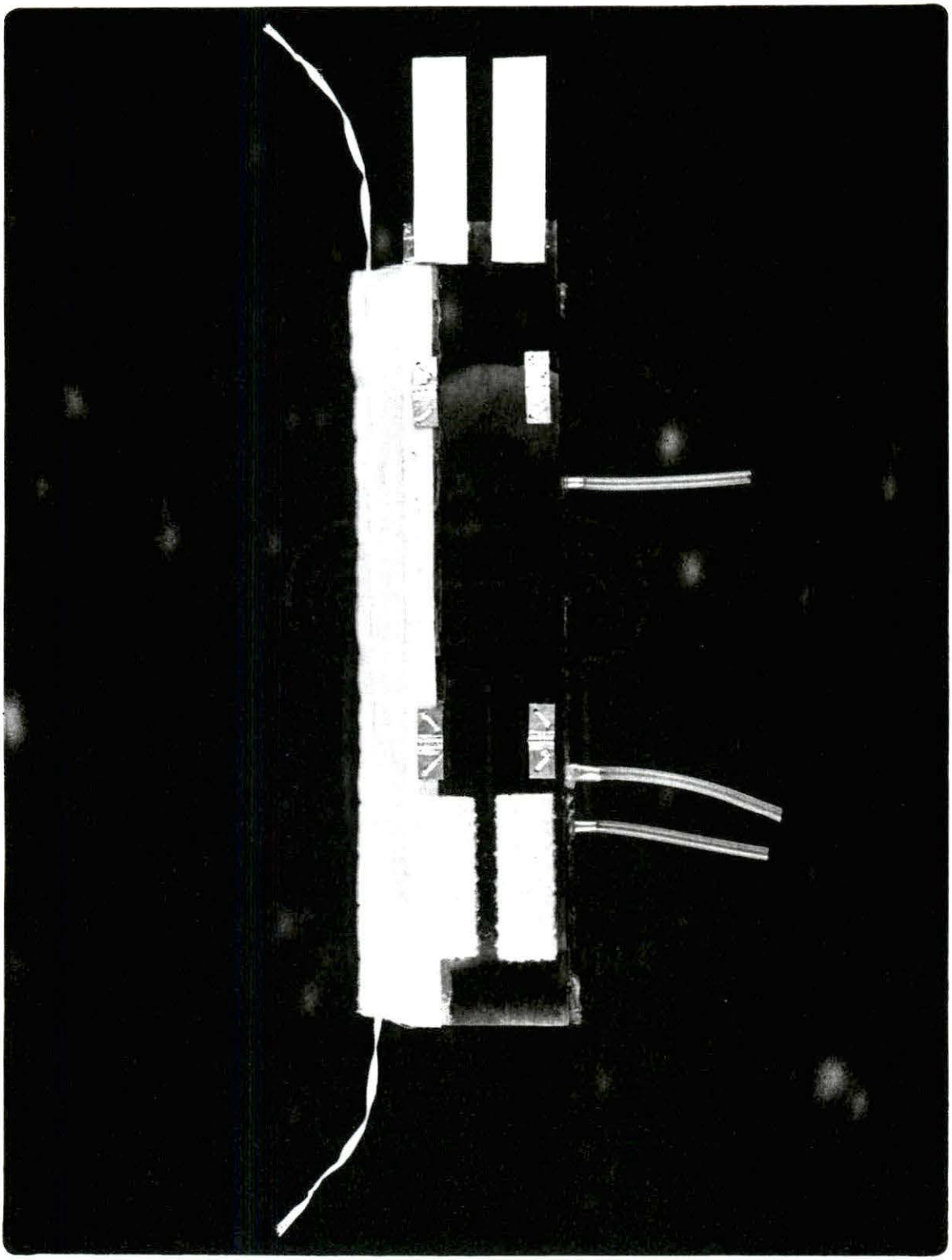
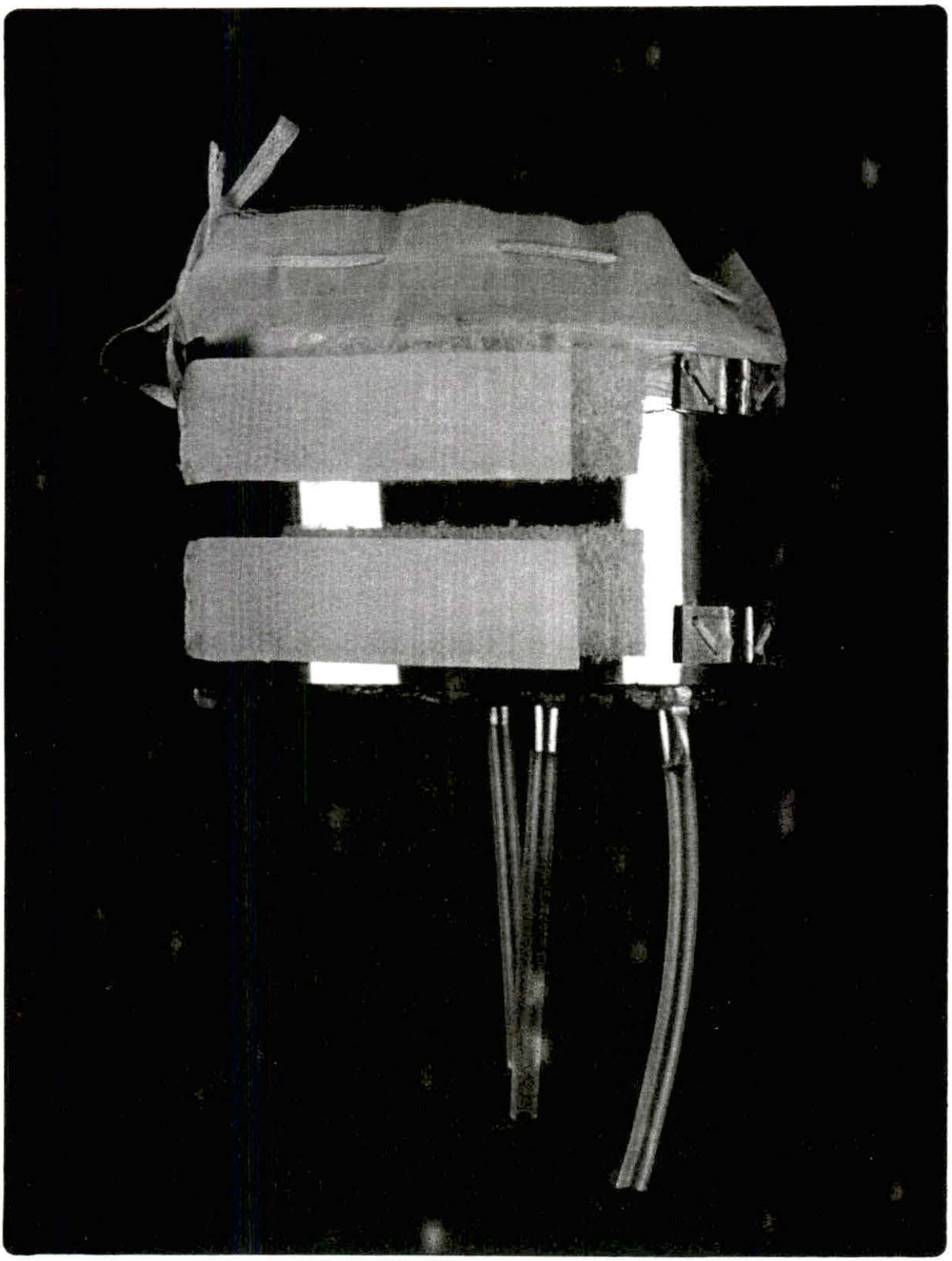


Figure 3.20. A photograph of the heart cuff in a cylindrical shape



When the heart cuff is wrapped around a heart, the outer shell forms a hollow cylinder. The ends of the cuff are overlapped and adjusted until the circumferences of the cuff and the maximum cross-section of the heart are equal. Velcro straps on each end of the outer shell are used to maintain the correct size. The bladder is inflated with saline until the bottom portion of the bladder fills the gap between the cylindrical outer shell and the approximately conical shaped heart. An illustration of the heart cuff positioned on a heart is shown in Figure 3.21. A photograph of an actual heart without and with the heart cuff in position is shown in Figure 3.22 and 3.23, respectively.

Two methods are used to prevent the cuff from being ejected from the heart. The first is a purse-string nylon "dome". A strip of nylon, with a piece of umbilical tape interwoven through it, is attached to the top of the outer shell. When the umbilical tape is pulled and tightened, the nylon wraps around the base of the heart, and the cuff is unable to slide off because the purse-string is secured in the atrioventricular groove. This method is illustrated in Figure 3.21.

The second method employs an aortic ring. The rigid ring, with straps attached, is placed around the ascending aorta. The straps are threaded through supports and then fastened to the heart cuff. The supports keep the straps above the base of the heart so that the tension produced in the straps doesn't occlude any of the blood vessels originating or terminating at the heart, namely the low pressure pulmonary artery and vena cava. This method is illustrated in Figure 3.24.

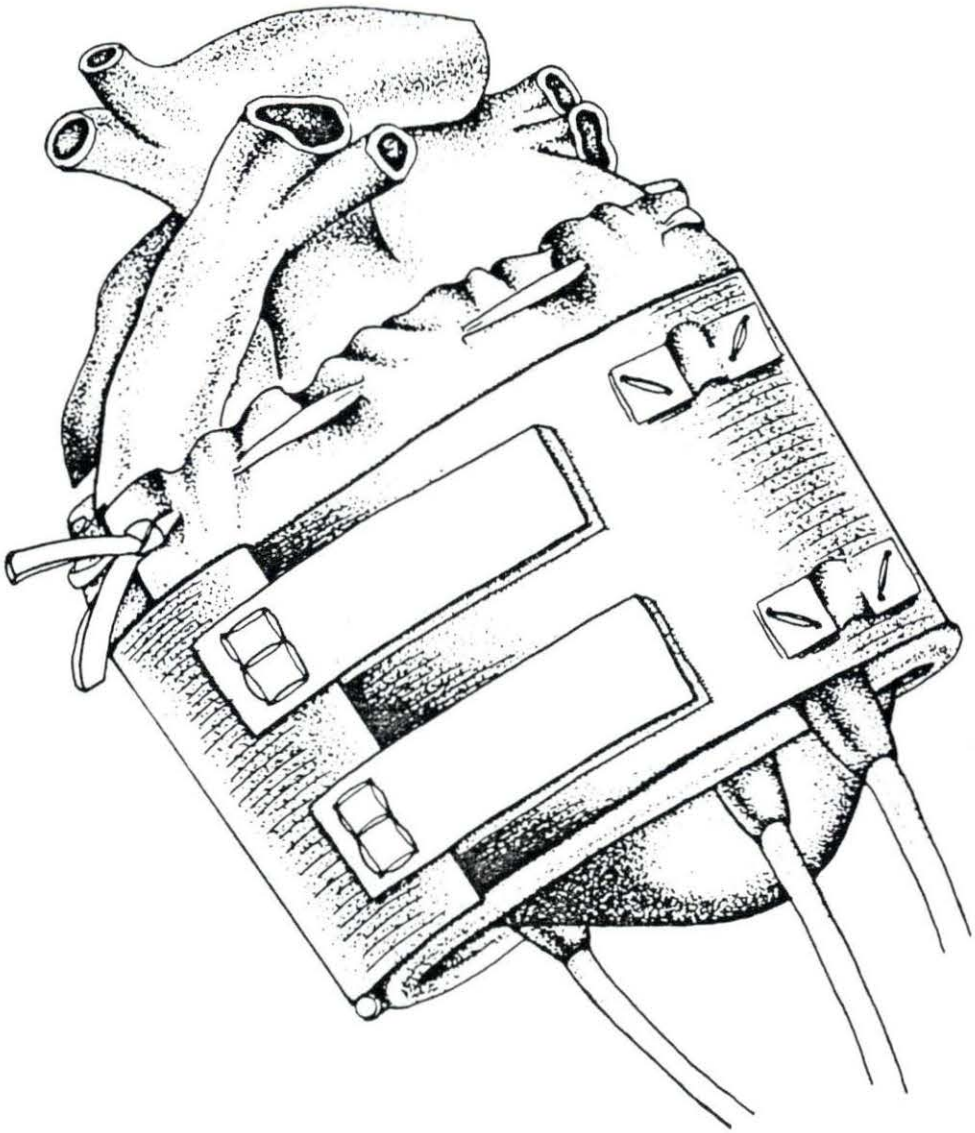


Figure 3.21. An illustration of the heart cuff in position on a heart using the purse-string nylon "dome" to keep it in place

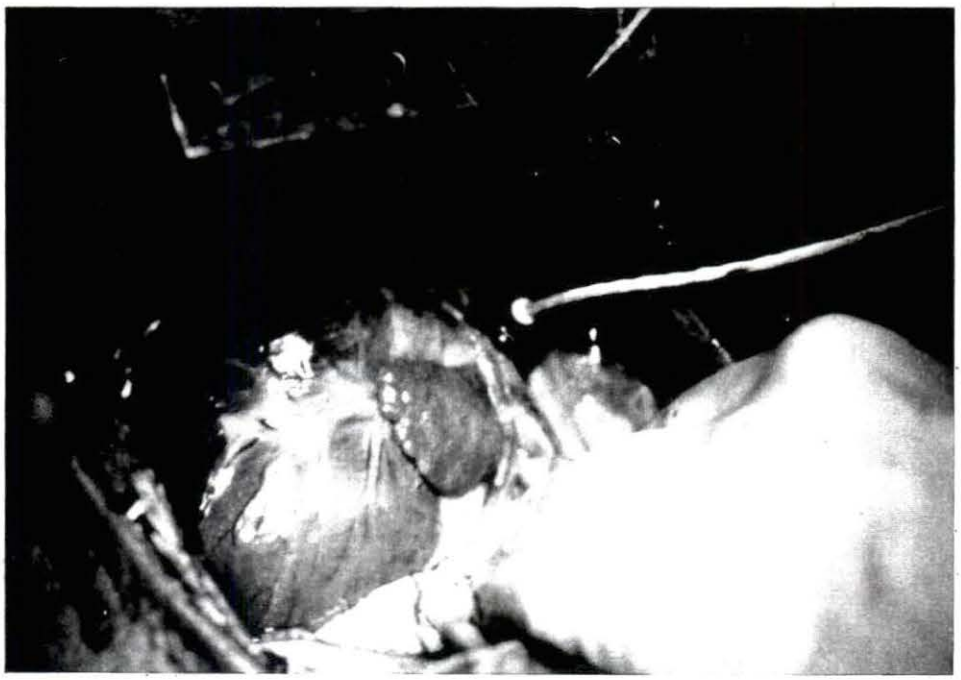


Figure 3.22. A photograph of a heart before the heart cuff was placed in position

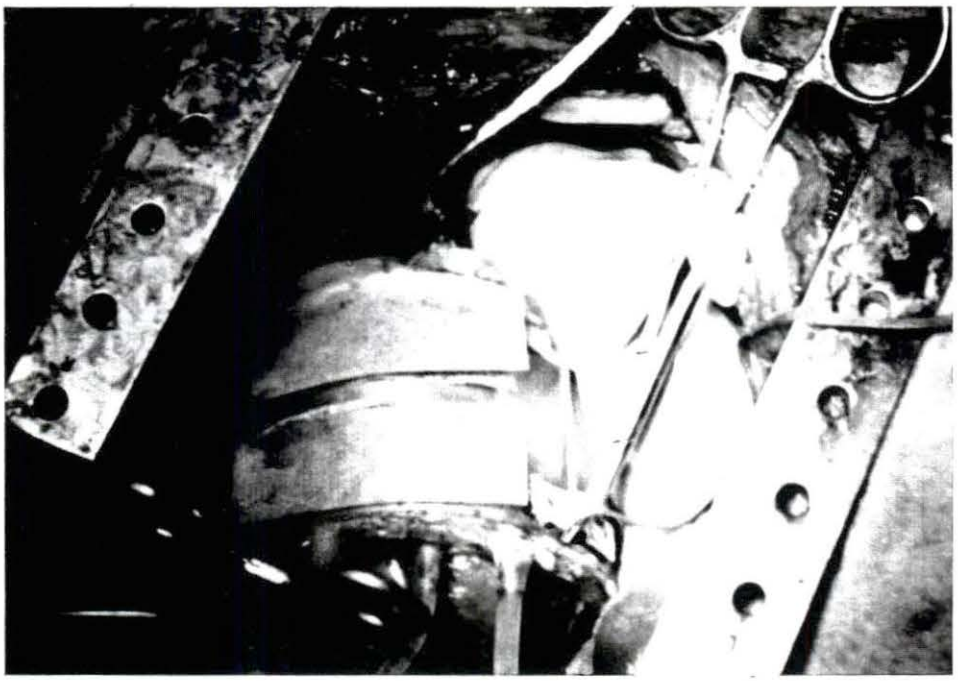


Figure 3.23. A photograph of the heart cuff in position on an actual heart

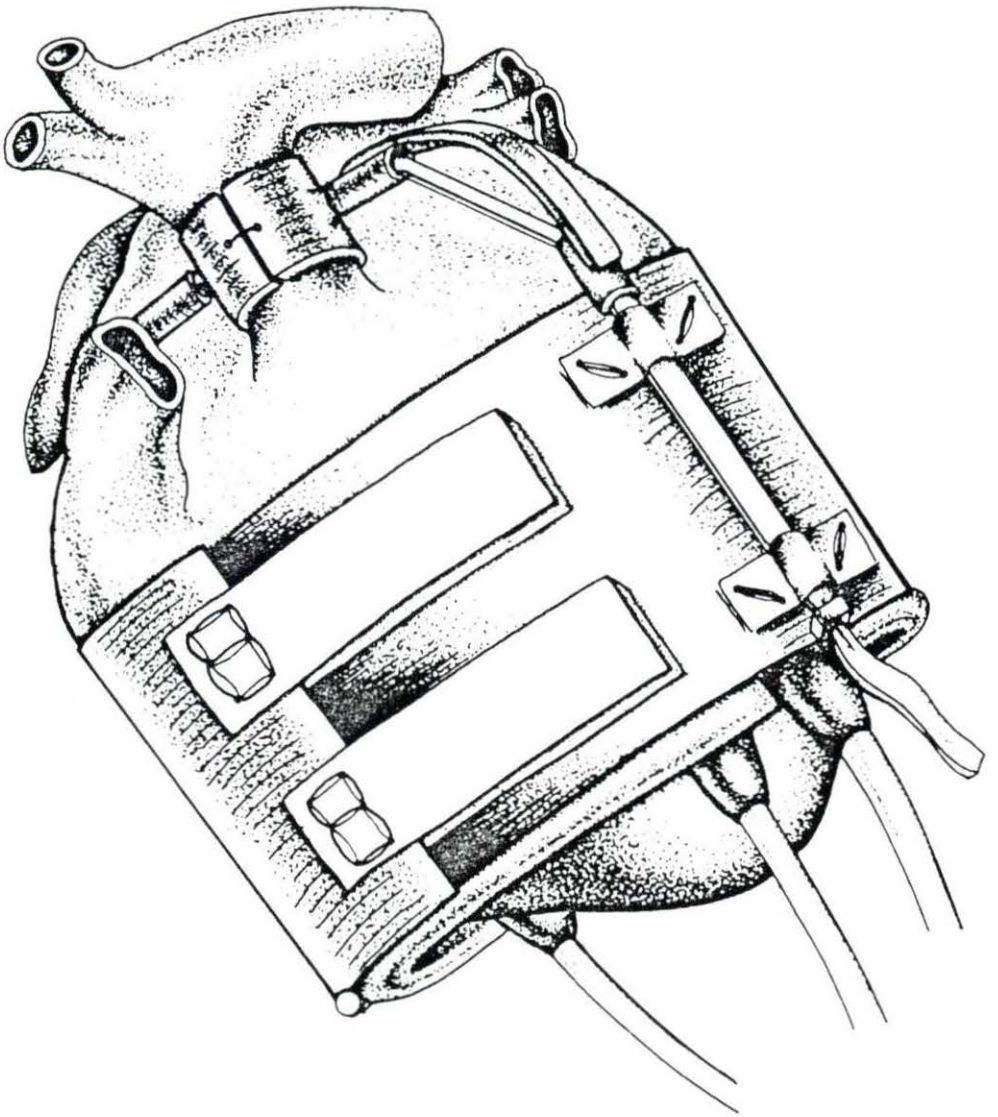


Figure 3.24. An illustration of the heart cuff in position on a heart using the aortic ring and straps with supports to keep it in place

CHAPTER 4. EVALUATION OF THE FINAL DESIGN

The heart cuff was evaluated through a series of six animal trials using mongrel dogs weighing between 19 and 25 kg. The objectives of the evaluation were:

1. to determine the effect the heart cuff had on a normal beating heart when the device was placed into position and the bladder inflated,
2. to further test the two methods developed to keep the heart cuff in proper position during assistance, and
3. to determine the performance of the device during assistance on a normal beating heart.

Three heart cuffs were fabricated with varying sizes to ensure a cuff was available to properly fit the heart during each animal trial. One heart cuff had a length of 25 cm and a width of 4.75 cm; the second had a length of 26 cm and a width of 5.0 cm; and the third had a length of 27 cm and a width of 5.5 cm. Since the circumference of the heart cuff is adjustable, these three cuffs would fit a variety of different sized hearts. The cuffs weigh approximately 60 grams (0.13 lb).

To evaluate each method for keeping the heart cuff in proper position on the heart, the purse-string nylon "dome" was used during the first three trials, and the aortic ring with the supported straps were used during the last three trials.

During the earlier tests, only the aortic pressure and flow were used to evaluate the performance of each prototype. For a better understanding of the performance of the device and the effect it had on the subject, several other parameters were recorded during this phase of the project. In addition to the aortic pressure and

flow, left ventricular pressure, central venous pressure, and the ECG were monitored. A CODAS data acquisition system with a Zenith personal computer was used to acquire the series of data. Using the post-acquisition functions on the CODAS system, the data was processed to determine the contractility of the left ventricle, the stroke volume of the flow pulse, the mean aortic flow, and the heart rate. The contractility of the left ventricle was calculated using the differential function which determined the derivative of the ventricular pressure waveform. The derivative was calculated to obtain the maximum slope of the left ventricular pressure during contraction. This was used as an indication of myocardial performance with and without assistance. The stroke volume was calculated using the integration function which determined the area under each aortic flow pulse. The mean flow was calculated using the moving average function which filtered the high frequency components of the signal. Finally, the heart rate was determined using the ECG.

Procedure

The same procedure was used as outlined in the surgical procedure section of the animal studies portion of this paper to expose the heart. A left thoracotomy in the fifth intercostal space was used to enter the thoracic cavity. The pericardium was incised ventrally to relieve the heart from the pericardium and to expose the epicardium.

The descending aorta was isolated and the 12 mm Transonic flow probe was placed around it to determine aortic flow. Since the ascending aorta was isolated for the placement of the aortic ring, the flow probe could not be placed at this

location, and thus, the total cardiac output minus coronary flow could not be obtained.

A surgical cut-down was performed on the left inner thigh to expose the femoral artery and vein. A Millar micro-tip pressure catheter was then introduced into the femoral artery, and the tip was positioned in the descending aorta to measure arterial pressure. A second Millar micro-tip pressure catheter was introduced into the femoral vein, and the tip was placed in the vena cava to measure central venous pressure.

A second surgical cut-down was performed on the left neck to expose the left carotid artery and jugular vein. A third Millar pressure catheter was introduced into the carotid artery, and the tip was placed inside the left ventricle to determine left ventricular pressure. The catheter with the ECG electrodes was introduced into the jugular vein, and the tip was positioned near the right ventricle until a good signal was obtained. Two of the four electrodes were connected to the balloon pump system to trigger the inflation of the balloons. The other two were connected to a monitor so the signal could be recorded.

To ensure there were no variations in the procedure from trial to trial which might affect the evaluation of data, the same protocol was followed precisely during each animal trial. After all the surgical cut-downs were performed, the heart was allowed to stabilize for 5 minutes. A control reading was then taken. The circumference of the heart was measured just below the atrioventricular groove, and a heart cuff was chosen to match the circumference. The cuff was then placed into position on the heart, and the bladder was left empty. After 5 minutes of recovery, another recording was taken. The bladder was then inflated using saline until either

the ventricular pressure dropped by 10% or a good fit was achieved. After the heart was allowed to stabilize for 5 minutes, another recording was taken. Immediately following, partial assistance every other beat was started and a recording was taken. Then, after 15 and 30 minutes of assisting every other beat, a recording was taken. After 30 minutes, the assistance was switched to every beat, and a recording was immediately taken. Then another recording was taken after 15 minutes of assisting every beat. The cuff was removed and the heart was allowed to stabilize for 5 minutes. A final recording was then taken.

This procedure was followed for each evaluation except animal trial #4. The heart rate was too fast, and the balloon pump was not properly triggering the inflation of the balloons during assistance every beat. Assistance every other beat was continued for 45 minutes and a recording was taken. This was done so each heart was assisted for the same amount of time.

Results

The values for systolic and diastolic aortic pressure, mean aortic flow, stroke volume, central venous pressure, left systolic ventricular pressure, peak contractility of the left ventricle, and heart rate were determined for each of the recorded time periods. Recordings of these waveforms obtained during the control period of animal trial #4 are shown in Figures 4.1 - 4.4 (each small time division is 0.2 seconds). Figure 4.1 shows a recording of the aortic flow with the mean aortic flow superimposed over it. Figure 4.2 shows a recording of the left ventricular pressure with the aortic pressure superimposed over it. Figure 4.3 shows a recording of the central venous pressure (top) and the ECG (bottom). Figure 4.4 shows a recording

of the accumulated area underneath each pulse of the aortic flow waveform (top) and the slope of the left ventricular pressure waveform (bottom). The stroke volume, the total area underneath a single aortic flow pulse, is obtained at the peak of the waveform showing total accumulation of the area. The slope of the left ventricular pressure waveform is the contractility of the left ventricle. Recordings of these waveforms during assistance every other beat after 30 minutes for animal trial #4 are shown in Figures 4.5 - 4.8. These recordings show the difference between the assisted beats and the unassisted beats for each parameter. The results of these parameters during each animal trial are shown in Tables 4.1 - 4.6.

In each trial, the aortic pressure, mean aortic flow, stroke volume, peak ventricular pressure, and peak contractility decreased, the central venous pressure increased, and the heart rate remained relatively constant when the heart cuff was placed on the heart and the bladder was pressurized. The stroke volume decreased by an average of 24.3 ± 6.9% from the control value when the cuff was placed into position, and it decreased by an average of 36.5 ± 5.4% from the control value when the bladder was subsequently inflated. The left systolic ventricular pressure decreased by an average of 11.3 ± 1.7% from the control value when the heart cuff was placed into position, and it decreased by an average of 20.7 ± 3.0% from the control value when the bladder was subsequently inflated. During partial assistance every other beat, aortic pressure, peak aortic flow, stroke volume, peak ventricular pressure, and peak contractility increased from the unassisted beats to the assisted beats. After 30 minutes of assisting every other beat, all of the parameters further increased except for central venous pressure and the heart rate which decreased slightly. The stroke volume increased by an average of 49.2 ± 13.8%, the peak ventricular pressure increased by an average of 28.0 ± 3.0%, and the mean aortic

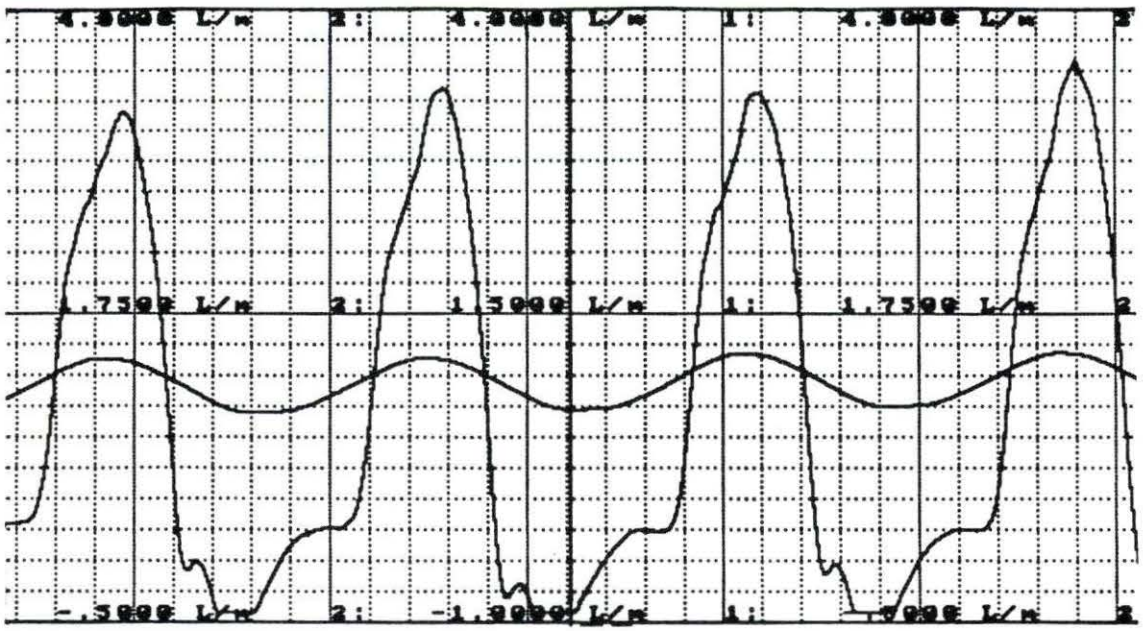


Figure 4.1. A recording of aortic flow and mean aortic flow before the heart cuff was placed on the heart during trial #4

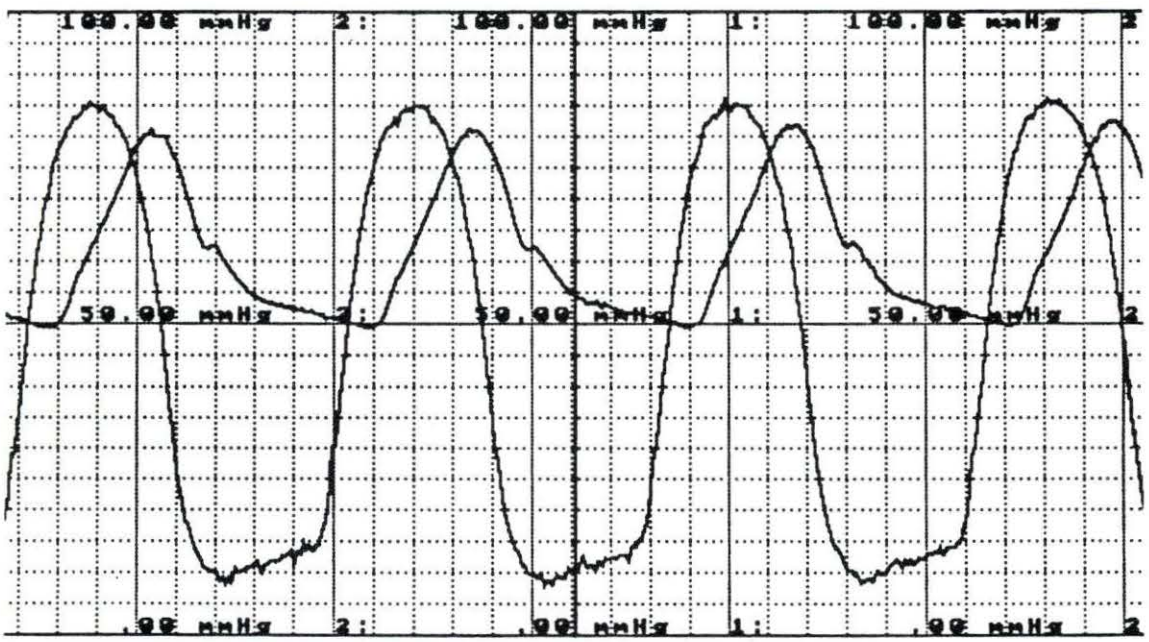


Figure 4.2. A recording of aortic and ventricular pressure before the heart cuff was placed on the heart during trial #4

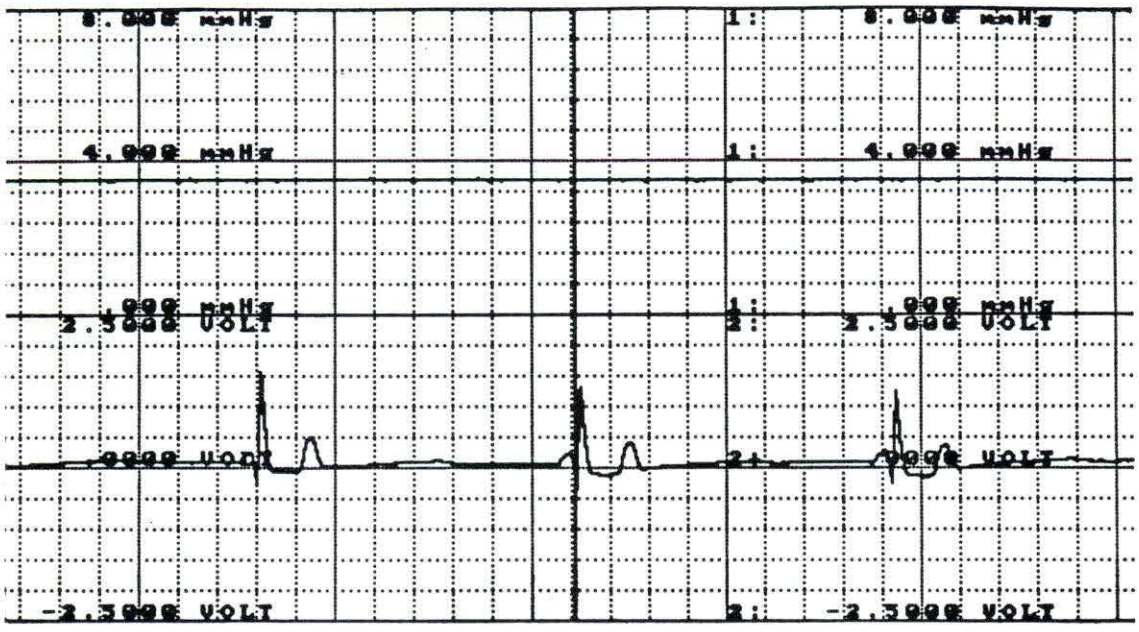


Figure 4.3. A recording of central venous pressure (top) and the ECG before the heart cuff was placed on the heart during trial #4

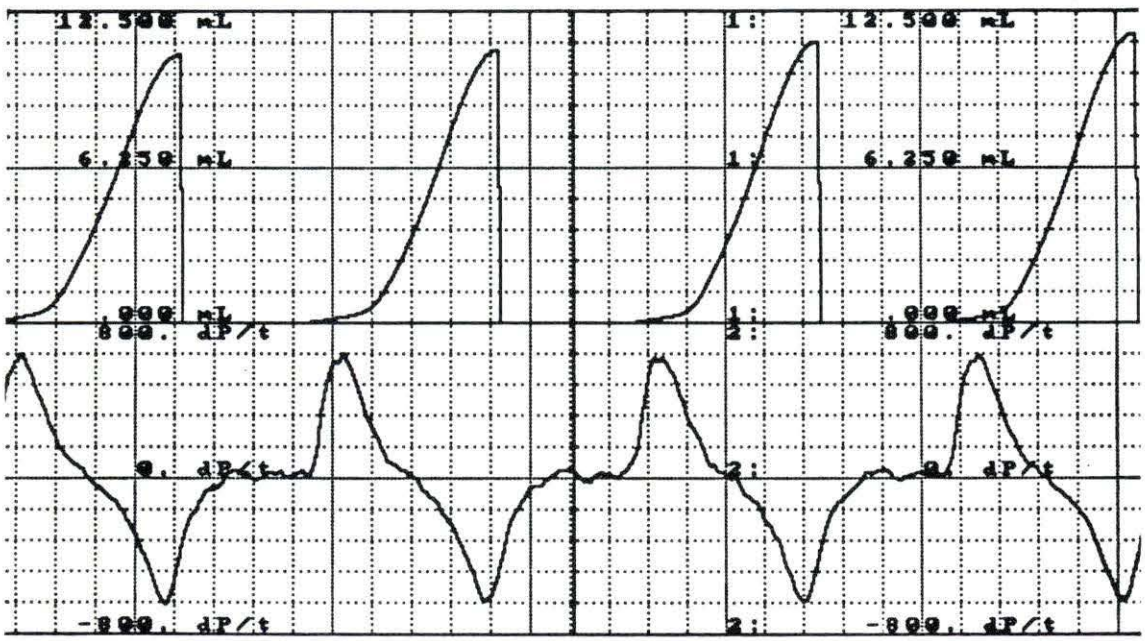


Figure 4.4. A recording showing stroke volume (top) and contractility before the heart cuff was placed on the heart during trial #4

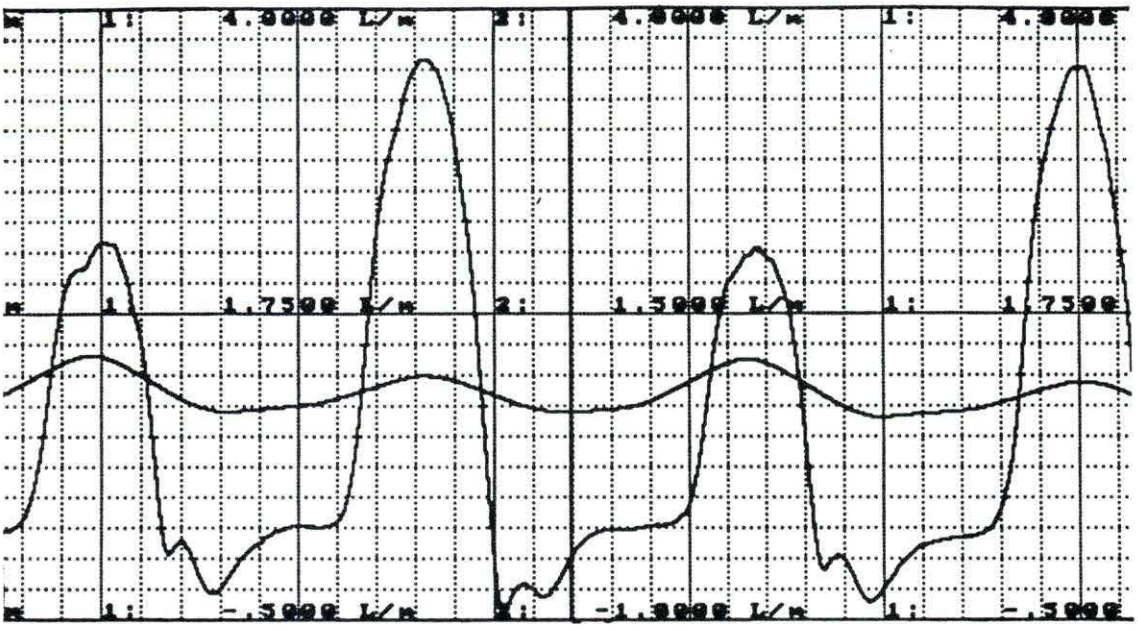


Figure 4.5. A recording of aortic flow and mean aortic flow after assisting every other beat for 30 min during trial #4

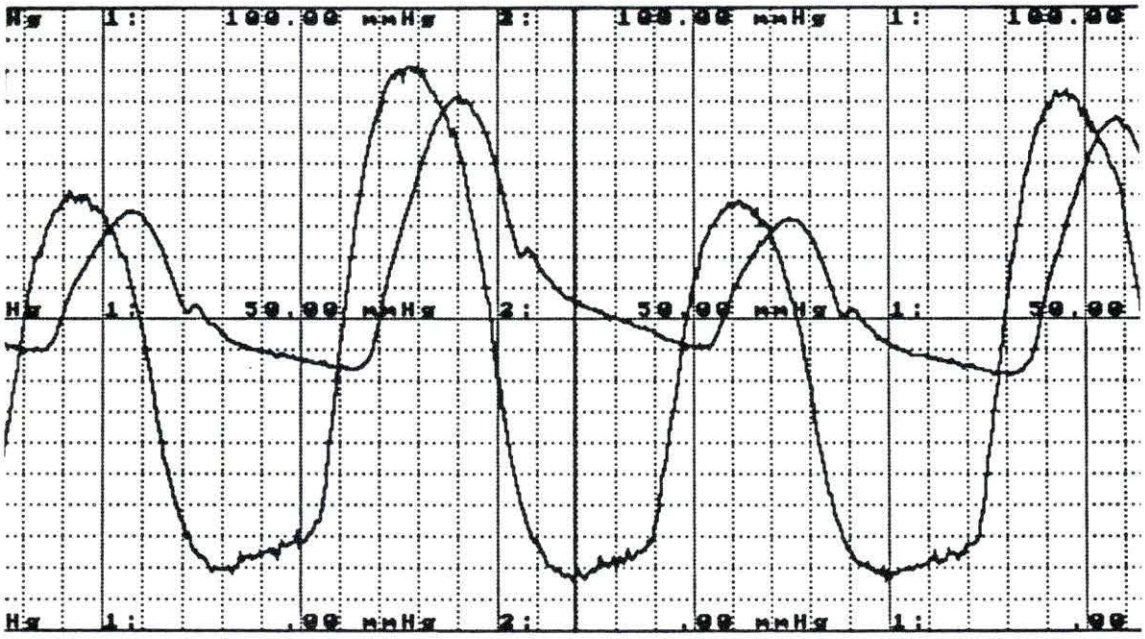


Figure 4.6. A recording of aortic and ventricular pressure after assisting every other beat for 30 min during trial #4

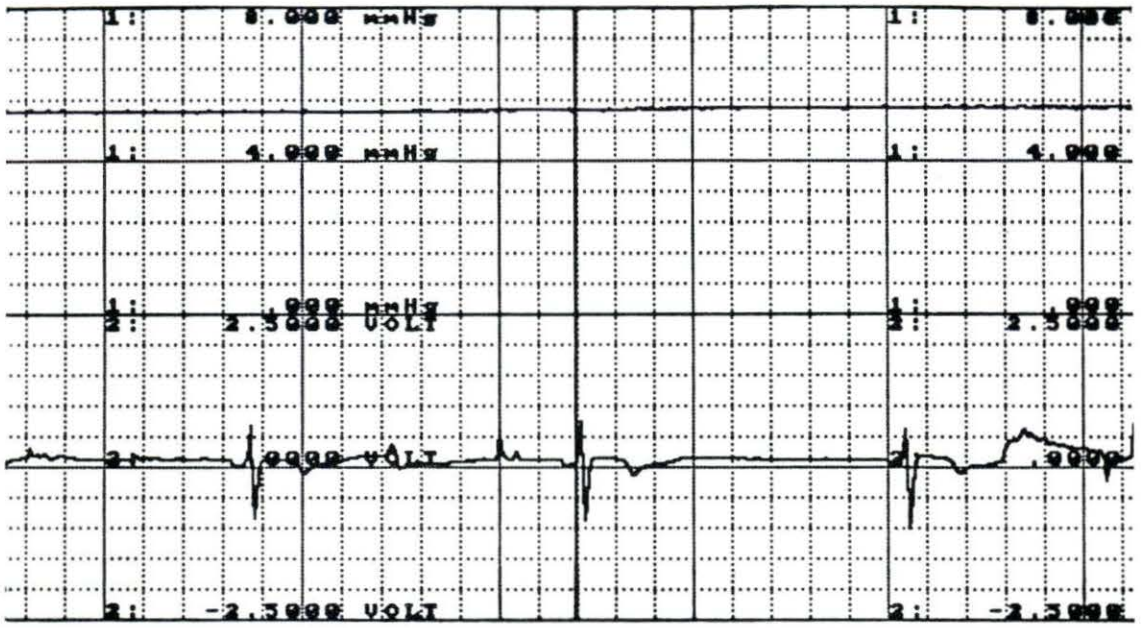


Figure 4.7. A recording of venous pressure (top) and the ECG after assisting every other beat for 30 min during trial #4

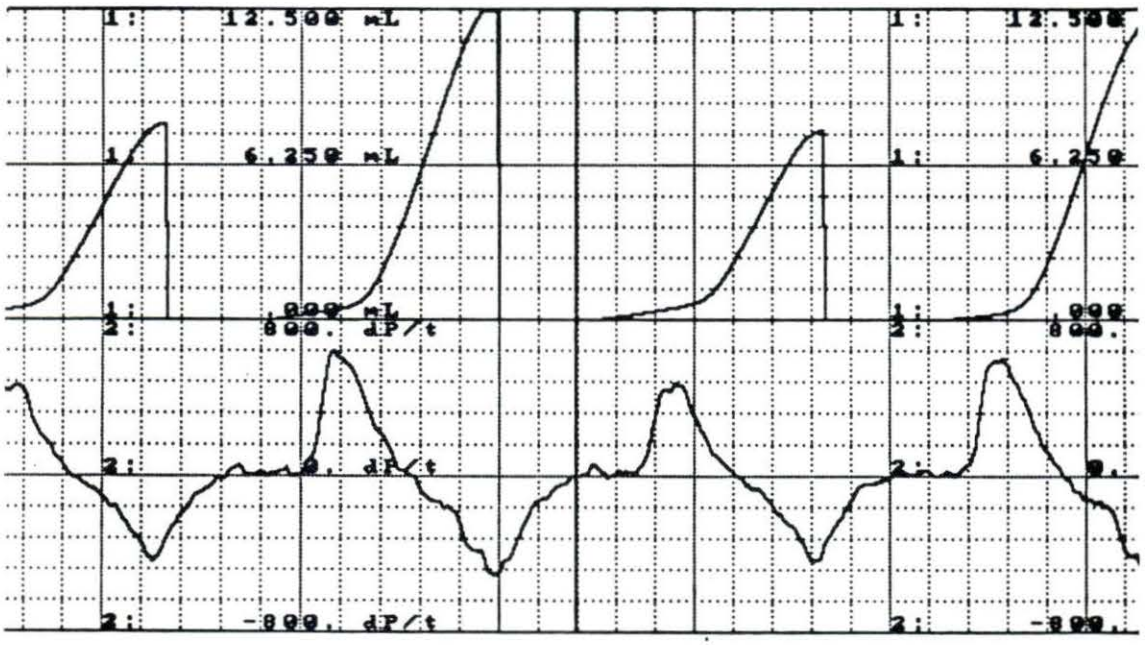


Figure 4.8. A recording showing stroke volume (top) and contractility after assisting every other beat for 30 min during trial #4

Table 4.1. Results of animal trial #1
Subject: 25 kg mongrel dog
Method of attachment: purse-string nylon "dome'

		Aortic Pressure (mm Hg)	Mean Aortic Flow (L/min)	Stroke Volume (mL)	Central Venous Pressure (mm Hg)	Left Systolic Ventricular Pressure (mm Hg)	Peak Contractility (mm Hg/s)	Heart Rate (bpm)
Control		86 / 50	2.0	14.9	5.3	78	1040	94
Cuff on: bladder not pressurized		74 /45	1.4	14.4	7.8	70	850	93
Cuff on: bladder pressurized		65 / 40	1.1	12.1	8.8	62	730	93
Assist every other beat: start	assist	71 / 41	1.1	12.6	9.0	70	780	90
	no assist	62 /42	1.1	11.6	9.0	62	670	90
Assist every other beat: 15 min	assist	78 / 44	1.2	12.7	8.5	74	790	89
	no assist	68 / 45	1.2	12.1	8.5	66	720	89
Assist every other beat: 30 min	assist	86 / 49	1.2	12.0	9.0	78	880	90
	no assist	76 / 49	1.2	12.0	9.0	70	800	90
Assist every beat: start		84 / 50	1.2	12.0	9.5	77	920	89
Assist every beat: 15 min		89 / 52	1.2	12.0	10.0	80	920	88
Assist every beat: 30 min		93 / 53	1.2	12.2	10.8	80	950	88
Post-assist: cuff off		88 / 53	1.1	11.5	11.1	77	870	89

Table 4.2. Results of animal trial #2
Subject: 25 kg mongrel dog
Method of attachment: purse-string nylon "dome'

		Aortic Pressure (mm Hg)	Mean Aortic Flow (L/min)	Stroke Volume (mL)	Central Venous Pressure (mm Hg)	Left Systolic Ventricular Pressure (mm Hg)	Peak Contractility (mm Hg/s)	Heart Rate (bpm)
Control		97 / 62	1.1	12.3	4.9	97	1050	96
Cuff on: bladder not pressurized		80 / 47	1.0	10.6	6.0	84	1010	104
Cuff on: bladder pressurized		75 / 44	0.9	9.4	6.9	80	970	106
Assist every other beat: start	assist	87 / 45	0.9	12.2	6.7	94	1150	103
	no assist	74 / 48	0.9	8.3	6.7	81	920	103
Assist every other beat: 15 min	assist	92 / 47	0.9	12.2	6.0	98	1150	102
	no assist	77 / 50	0.9	8.4	6.0	83	1000	102
Assist every other beat: 30 min	assist	94 / 48	1.0	12.2	5.0	100	1230	100
	no assist	79 / 51	1.0	8.8	5.0	85	1060	100
Assist every beat: start		90 / 48	1.0	11.5	5.3	98	1220	100
Assist every beat: 15 min		91 / 46	1.1	12.2	5.4	101	1390	93
Assist every beat: 30 min		79 / 47	1.2	14.0	2.8	88	1120	95

Table 4.3. Results of animal trial #3
Subject: 19 kg mongrel dog
Method of attachment: purse-string nylon "dome'

		Aortic Pressure (mm Hg)	Mean Aortic Flow (L/min)	Stroke Volume (mL)	Central Venous Pressure (mm Hg)	Left Systolic Ventricular Pressure (mm Hg)	Peak Contractility (mm Hg/s)	Heart Rate (bpm)
Control		94 / 63	1.0	11.2	4.0	93	930	95
Cuff on: bladder not pressurized		78 / 52	0.7	7.7	6.6	79	750	96
Cuff on: bladder pressurized		69 / 45	0.5	6.0	7.9	68	620	92
Assist every other beat: start	assist	78 / 46	0.5	8.0	8.1	78	660	92
	no assist	64 / 48	0.5	4.9	8.1	65	590	92
Assist every other beat: 15 min	assist	87 / 49	0.6	9.0	7.9	84	740	96
	no assist	70 / 53	0.6	4.4	7.9	70	610	96
Assist every other beat: 30 min	assist	85 / 50	0.6	8.3	8.3	83	770	96
	no assist	70 / 53	0.6	4.9	8.3	71	630	96
Assist every beat: start		88 / 57	0.6	7.2	8.3	85	760	95
Assist every beat: 15 min		87 / 56	0.6	7.2	7.7	85	830	93
Assist every beat: 30 min		87 / 54	0.7	7.8	7.0	87	750	91
Post-assist: cuff off		93 / 58	0.7	8.5	4.0	92	770	95

Table 4.4. Results of animal trial #4

Subject: 19 kg mongrel dog

Method of attachment: aortic ring with straps

		Aortic Pressure (mm Hg)	Mean Aortic Flow (L/min)	Stroke Volume (mL)	Central Venous Pressure (mm Hg)	Left Systolic Ventricular Pressure (mm Hg)	Peak Contractility (mm Hg/s)	Heart Rate (bpm)
Control		81 / 50	1.0	11.0	3.5	77	650	92
Cuff on: bladder not pressurized		73 / 43	0.8	8.5	5.9	70	580	94
Cuff on: bladder pressurized		69 / 42	0.7	7.2	7.0	65	550	95
Assist every other	assist	80 / 40	0.9	11.9	7.6	79	620	88
beat: start	no assist	62 / 43	0.9	6.8	7.6	60	470	88
Assist every other	assist	83 / 40	0.9	13.1	5.9	80	620	89
beat: 15 min	no assist	63 / 45	0.9	7.5	5.9	60	450	89
Assist every other	assist	81 / 40	0.9	12.5	5.5	79	610	89
beat: 30 min	no assist	65 / 44	0.9	8.1	5.5	62	460	89
Assist every other	assist	83 / 41	0.9	11.9	5.2	80	600	89
beat: 45 min	no assist	69 / 45	0.9	7.7	5.2	63	460	89
Post-assist: cuff off		71 / 40	0.9	10.6	3.1	63	490	84

Table 4.5. Results of animal trial #5
Subject: 24 kg mongrel dog
Method of attachment: aortic ring with straps

		Aortic Pressure (mm Hg)	Mean Aortic Flow (L/min)	Stroke Volume (mL)	Central Venous Pressure (mm Hg)	Left Systolic Ventricular Pressure (mm Hg)	Peak Contractility (mm Hg/s)	Heart Rate (bpm)
Control		58 / 40	0.9	10.7	4.4	65	510	87
Cuff on: bladder not pressurized		62 / 40	0.4	5.0	9.2	62	520	85
Cuff on: bladder pressurized		57 / 35	0.5	4.9	9.0	58	490	84
Assist every other beat: start	assist	69 / 35	0.6	8.4	8.7	75	470	83
	no assist	54 / 39	0.6	3.8	8.7	55	440	83
Assist every other beat: 15 min	assist	76 / 37	0.8	10.3	8.7	82	560	82
	no assist	61 / 42	0.8	6.3	8.7	62	520	82
Assist every other beat: 30 min	assist	74 / 36	0.7	9.6	8.5	82	540	81
	no assist	62 / 40	0.7	5.6	8.5	62	520	81
Assist every beat: start		71 / 40	0.7	7.7	9.1	75	520	81
Assist every beat: 15 min		84 / 44	0.6	7.8	9.4	86	590	76

Table 4.6. Results of animal trial #6
Subject: 20 kg mongrel dog
Method of attachment: aortic ring with straps

		Aortic Pressure (mm Hg)	Mean Aortic Flow (L/min)	Stroke Volume (mL)	Central Venous Pressure (mm Hg)	Left Systolic Ventricular Pressure (mm Hg)	Peak Contractility (mm Hg/s)	Heart Rate (bpm)
Control		97 / 60	1.0	11.4	7.4	94	810	99
Cuff on: bladder not pressurized		80 / 50	0.8	8.9	5.9	79	720	101
Cuff on: bladder pressurized		64 / 42	0.6	6.7	7.1	65	580	101
Assist every other beat: start	assist	83 / 43	0.7	10.0	7.0	87	700	102
	no assist	60 / 50	0.7	4.8	7.0	65	580	102
Assist every other beat: 15 min	assist	82 / 43	0.6	10.4	7.0	84	700	101
	no assist	62 / 47	0.6	4.6	7.0	62	530	101
Assist every other beat: 30 min	assist	82 / 44	0.7	10.0	6.3	86	680	100
	no assist	65 / 49	0.7	4.9	6.3	64	550	100
Assist every beat: start		75 / 45	0.6	7.2	7.0	75	640	101
Assist every beat: 15 min		83 / 47	0.7	8.5	6.5	80	620	99
Post-assist: cuff off		80 / 46	0.9	9.9	5.6	74	640	98

flow increased by an average of 19.7 ± 6.3% from the values obtained with the cuff in position and the bladder inflated after 30 minutes of assisting every other beat. During assistance every beat, aortic pressure, peak aortic flow, stroke volume, left systolic ventricular pressure, and peak contractility were maintained at the values obtained during assistance every other beat except for stroke volume which decreased slightly. After assisting every beat for 15 minutes, the stroke volume increased by an average of 29.4 ± 9.2%, the ventricular pressure increased by an average of 30.2 ± 4.6%, and the mean aortic flow increased by an average of 19.5 ± 4.4% from the value obtained after the cuff was in position and the bladder was inflated. The average percent change of the stroke volume and peak ventricular pressure at selected time intervals is shown in Table 4.7. In this table, "no assist" refers to the alternate beats in which no assist was given.

During some of the unassisted beats, the stroke volume and left systolic ventricular pressure decreased below the values obtained after the cuff was placed into position and the bladder inflated. Thus, a negative percentage is presented in Table 4.7. Since during the assisted beat the normal heart was ejecting an abnormally large amount of blood from the ventricles, the end systolic volume was low after the assisted beats. Subsequent ventricular filling before the following unassisted beat was not sufficient enough to obtain a normal end diastolic volume. Thus, the stroke volume and the systolic ventricular pressure of the unassisted beat was lower than normal. Following the low ejection fraction of the unassisted beat, the ventricles were able to adequately fill between assisted beats, and a larger quantity of blood was ejected again during the subsequent assisted beat. This was further demonstrated when the assistance was switched from every other beat to

Table 4.7. The average percent change of stroke volume and peak ventricular pressure compared to the values obtained after the cuff was in position and the bladder inflated

		Stroke Volume (%)	Left Systolic Ventricular Pressure (%)
Assist every other beat: start	assist	+42.0 ± 10.1	+21.8 ± 3.4
	no assist	-14.8 ± 3.9	-2.7 ± 1.3
Assist every other beat: 30 min	assist	+49.2 ± 13.8	+28.0 ± 3.0
	no assist	-4.2 ± 6.7	+3.8 ± 2.7
Assist every beat: start		+21.2 ± 9.6	+23.2 ± 2.3
Assist every beat: 15 min		+29.4 ± 9.2	+30.2 ± 4.6

every beat. Since the heart was being compressed by the assist device every beat, the heart had an abnormally high ejection fraction each beat, instead of every other beat. Since the ventricles were not able to adequately fill, the stroke volume was lower than during assistance every other beat.

Both the purse-string nylon "dome" and the aortic ring with supportive straps successfully kept the heart cuff in position throughout the assistance periods. The drop in pressure and flow which occurred when the device was placed on the heart was the same for both methods of attachment. In addition, the amount of assistance during the use of each method was the same. Therefore, one method did not appear to interfere with normal heart function or assistance more than the other.

CHAPTER 5. CONCLUSIONS

A direct mechanical ventricular assist device (heart cuff) was designed and developed which shows promise and demonstrates the feasibility of this type of device. The heart cuff can be quickly and easily slipped into position. The circumference of the outer shell is adjustable to match the size of a heart, and the bladder can be inflated to approximately fit the shape of the ventricles. Thus, only a few models are needed to apply the heart cuff to several different sized and shaped hearts.

Through the use of the purse-string nylon "dome" and/or the aortic ring with supportive straps, the heart cuff is prevented from being ejected from a heart. Since the bladder acts as a wedge between the cylindrical outer shell and the irregular shaped heart, the heart cuff is stabilized during assistance. During the evaluation of the device, these methods successfully kept the heart cuff in position for periods of over 1 hour with no evidence that the cuff was being ejected. Longer periods of assistance were not attempted in this study.

The device, when placed on the heart, does effect ventricular filling and contractility. The left systolic ventricular pressure dropped by an average of 20.7 ± 3.0% and the stroke volume dropped by an average of 36.5 ± 5.4% when the cuff was placed on the heart and the bladder was pressurized. This problem can perhaps be partially alleviated through the use of other materials and further refinement of the design. Since the heart cuff adds additional weight and bulkiness to the heart, the drop in pressure is always likely to be present. Since most previous attempts at

mechanical assistance were conducted on fibrillating hearts, no information is available to compare the results of the present study with those of other assist devices used on a normal heart.

The balloon pump was only capable of producing a 17 mL balloon volume. Thus, the amount of force delivered to the heart was limited. With a larger balloon volume, the heart can be further compressed, and larger pressures and flows can be achieved. This increase in balloon volume would certainly be necessary for assistance of human hearts.

Even though the balloon volume was limited, the heart cuff successfully increased aortic and ventricular pressure, stroke volume, aortic flow, and heart contractility during partial assistance on a normal beating heart. The left systolic ventricular pressure increased by 28.0 ± 3.0%, the stroke volume increased by 49.2 ± 13.8%, and the mean aortic flow increased by 19.7 ± 6.3% after assisting every other beat for 30 minutes when the percentages are averaged for the six dogs tested. This demonstrates that the heart cuff successfully supplemented the hemodynamics on a normal beating heart.

Previous studies have shown that direct mechanical assistance does not cause enough tissue damage to limit its applicability. Although a study was not performed to determine if the heart cuff caused tissue damage during assistance, two "localized" balloons were used to keep any damage which might occur to a limited area. This is believed to be better than compressing the entire heart, and causing potential damage over the entire surface area.

The specified design criteria were satisfied except for the requirement that the device not interfere with heart contractility and ventricular filling. This was only partially satisfied. However, with further development and testing, it is believed

that this problem can be overcome. The direct mechanical ventricular assist device described in this thesis proved to be feasible, and continued research is warranted.

CHAPTER 6. RECOMMENDATION FOR FURTHER RESEARCH

The heart cuff was initially evaluated on a normal beating heart. However, the heart cuff is intended to be used on a "weakened" heart. Thus, further research is necessary to determine the performance of the device on a heart with "stunned" myocardium. "Stunned" myocardium has a temporary decrease of function which can result in a decrease in cardiac output depending on the amount of effected tissue, but it is capable of fully recovering without any permanent damage. In order to properly evaluate the heart cuff on a "weakened" heart, a reliable and reproducible model of "stunned" myocardium must be achieved. Otherwise, the results of each evaluation will vary, and an accurate conclusion of the performance of the heart cuff could not be drawn.

"Stunned" myocardium can be obtained in dogs by causing a temporary "regional" or "global" ischemia. Temporary regional ischemia can be induced by occluding either the left circumflex coronary artery or the left anterior descending coronary artery for a prescribed time period. This will prevent blood flow to a localized area of the left ventricle. Temporary global ischemia can be induced by fibrillating the heart for a prescribed time period with subsequent defibrillation to regain heart activity. This will prevent blood flow to the entire left and right ventricle. The temporary loss of oxygen and nutrient supply "stuns" the sections of the myocardium effected by the ischemia, and the function of these sections is sacrificed. This can be demonstrated by observing the segmental shortening of the ischemic myocardial fibers. Either the ischemic myocardial fibers are unable to contract to their normal

diastolic length or they are unable to contract at all and dilate during systole. With a portion or all of the myocardium "stunned", the performance of the heart will be sacrificed, and the cardiac output and arterial pressures will decrease. This condition would be similar to a heart which has just been restarted after open heart surgery and is too weak to maintain adequate cardiac output.

Segmental shortening can be observed using a sonomicrometer with a pair of 5-MHz hemispherical piezoelectric ultrasonic crystals. The two transducer leads can be implanted into the subendocardium of the ischemic zone, approximately 10 mm apart and orientated parallel to the minor axis of the heart. The distance between the two transducer leads is related to the time it takes for an ultrasonic pulse to travel from one crystal to another. When the heart contracts during systole, the two crystals will move closer together. When the heart relaxes during diastole, the two crystals will move farther apart. Thus, the segmental shortening can be determined by continuously recording the distance between the two leads throughout a cardiac cycle.

The objectives of circulatory assistance are to decrease the workload of the failing heart while maintaining adequate peripheral blood flow (Spencer et al.). This will decrease the oxygen requirement of the myocardium and it will increase the perfusion of vital organs, such as the liver and kidneys. With the temporary decrease in the workload of the heart, heart function may improve, and with the increase in cardiac output, the general body metabolism may improve.

The heart cuff can be evaluated by determining how well it satisfies the objectives of assisted circulation. The increase in peripheral blood flow during assistance can be observed by recording several hemodynamic parameters, namely aortic pressure and flow, ventricular pressure, central venous pressure, and right

atrial pressure. The function of the heart can be observed by recording the changes in segmental shortening and the left ventricular contractility (slope of left ventricular pressure waveform) during assistance. With the workload of the heart decreased during mechanical assistance, the ischemic myocardial fibers should be able to contract to a shorter length than without assistance. In addition, the time required to fully recover can be compared to a control group in which no assistance was given. Ideally, the recovery should be quicker for the assisted hearts. This type of testing will more closely simulate the proposed usage of the heart cuff.

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APPENDIX: RECORDINGS FOR ANIMAL TRIAL #2

The complete recordings obtained during each time period of animal trial #2 showing the waveforms for aortic flow and mean aortic flow, ventricular and aortic pressure, central venous pressure and the ECG, and contractility and stroke volume are presented in Figures 8.1 - 8.36. Each small division on the horizontal scale is 0.2 seconds. A similar set of recordings is available for the other five dog trials used in this test program.

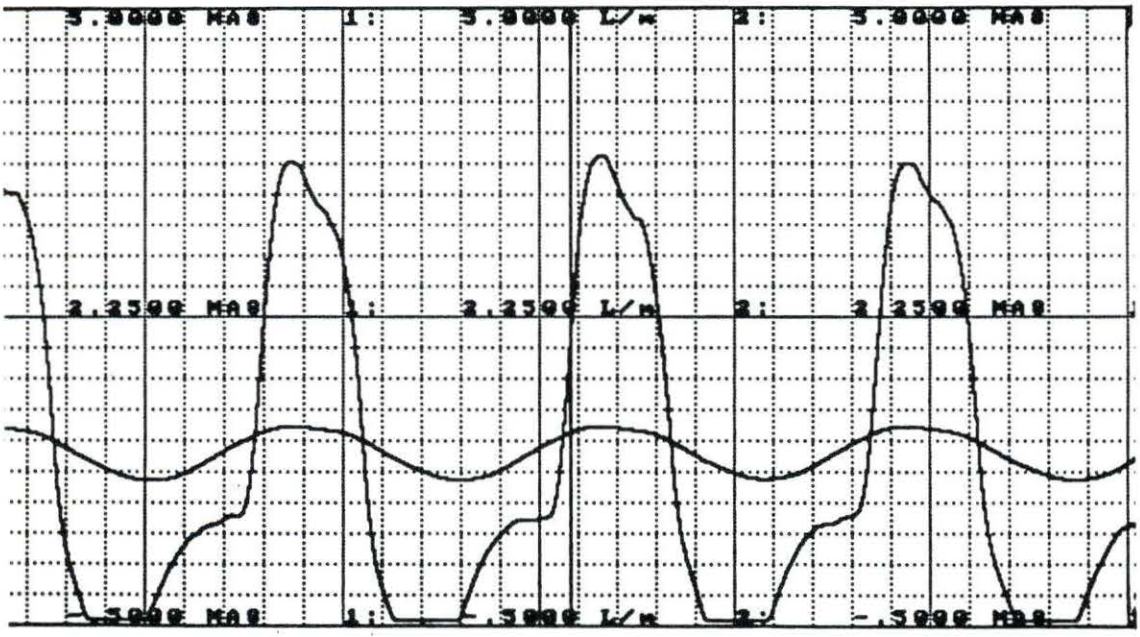

Figure A.1. A recording of aortic flow and mean aortic flow before the heart cuff was placed on the heart during trial #2

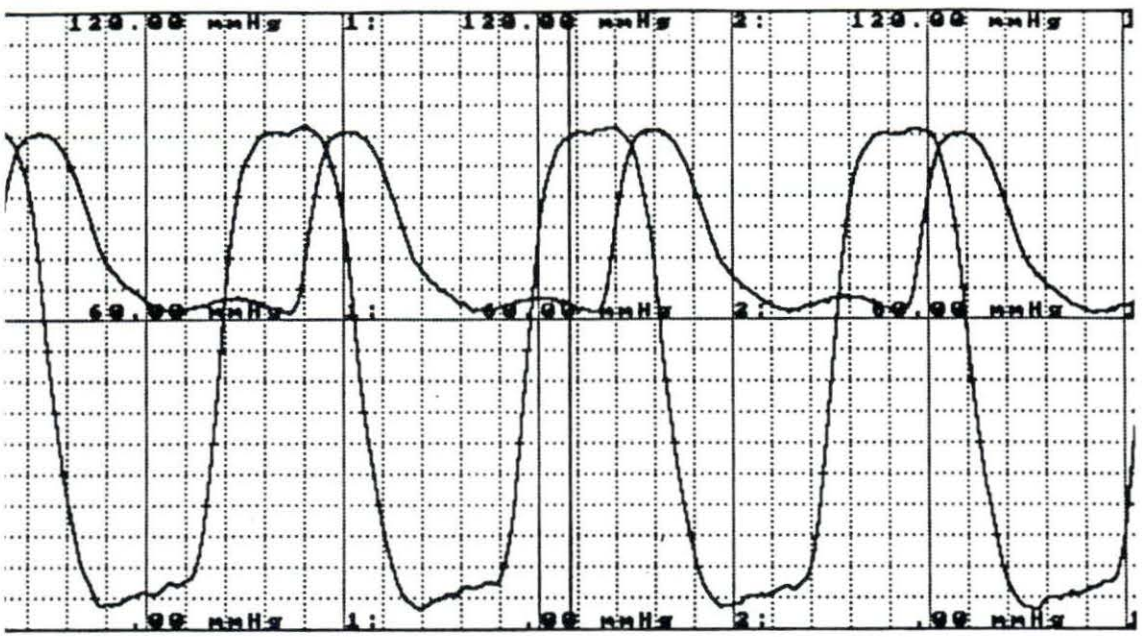

Figure A.2. A recording of aortic and ventricular pressure before the heart cuff was placed on the heart during trial #2

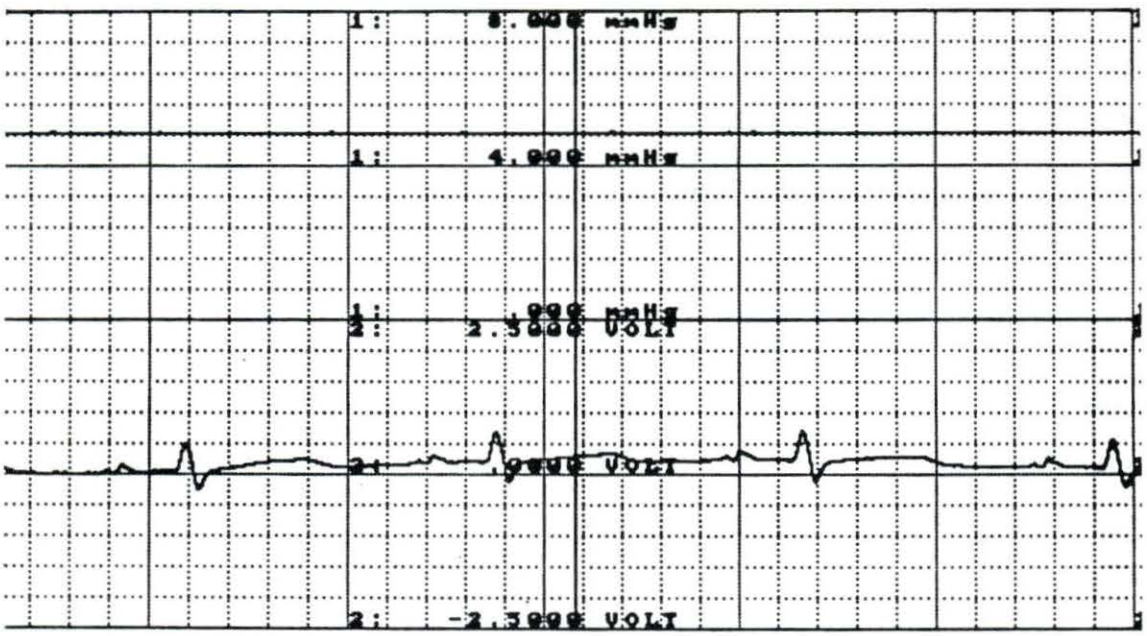

Figure A.3. A recording of central venous pressure and the ECG before the heart cuff was placed on the heart during trial #2

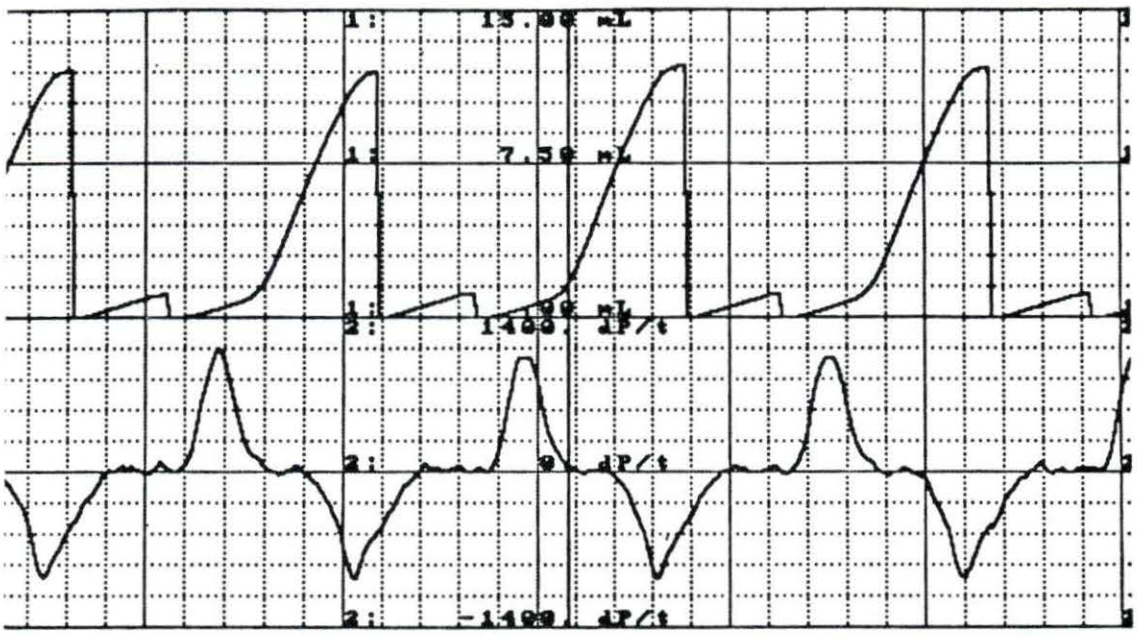

Figure A.4. A recording of stroke volume and contractility before the heart cuff was placed on the heart during trial #2

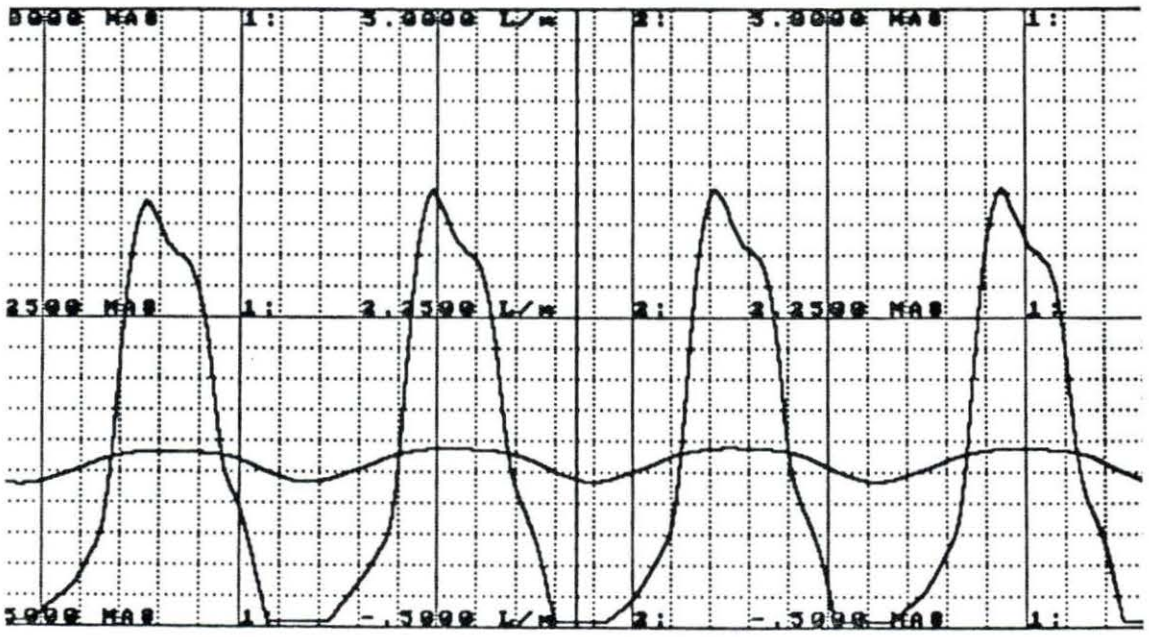

Figure A.5. A recording of aortic flow and mean aortic flow after the heart cuff (bladder empty) was placed on the heart during trial #2

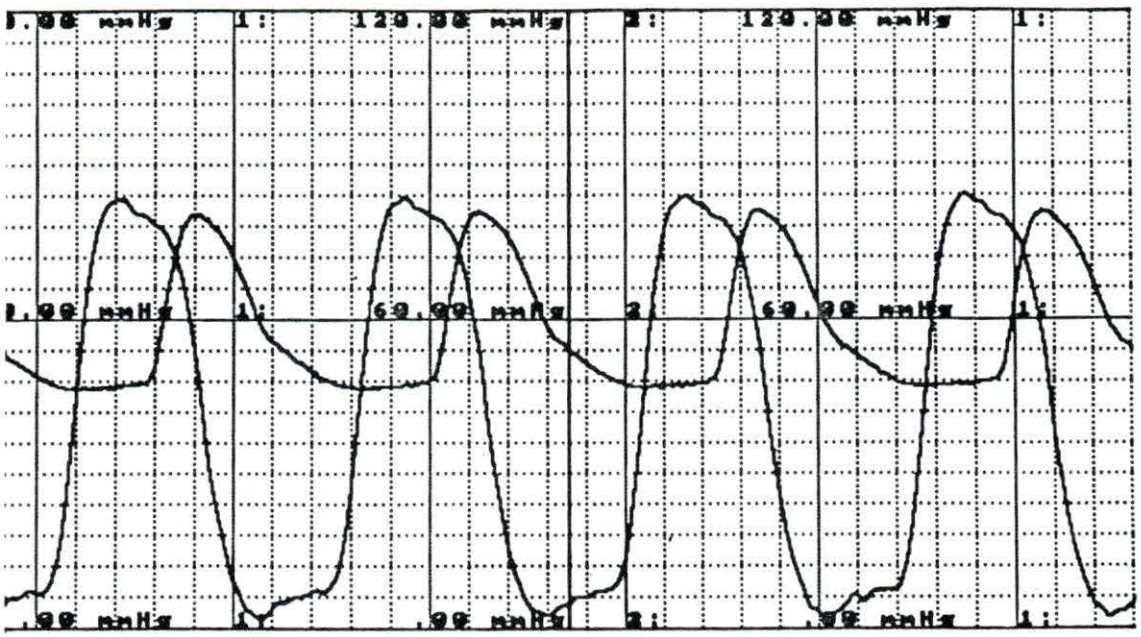

Figure A.6. A recording of aortic and ventricular pressure after the heart cuff (bladder empty) was placed on the heart during trial #2

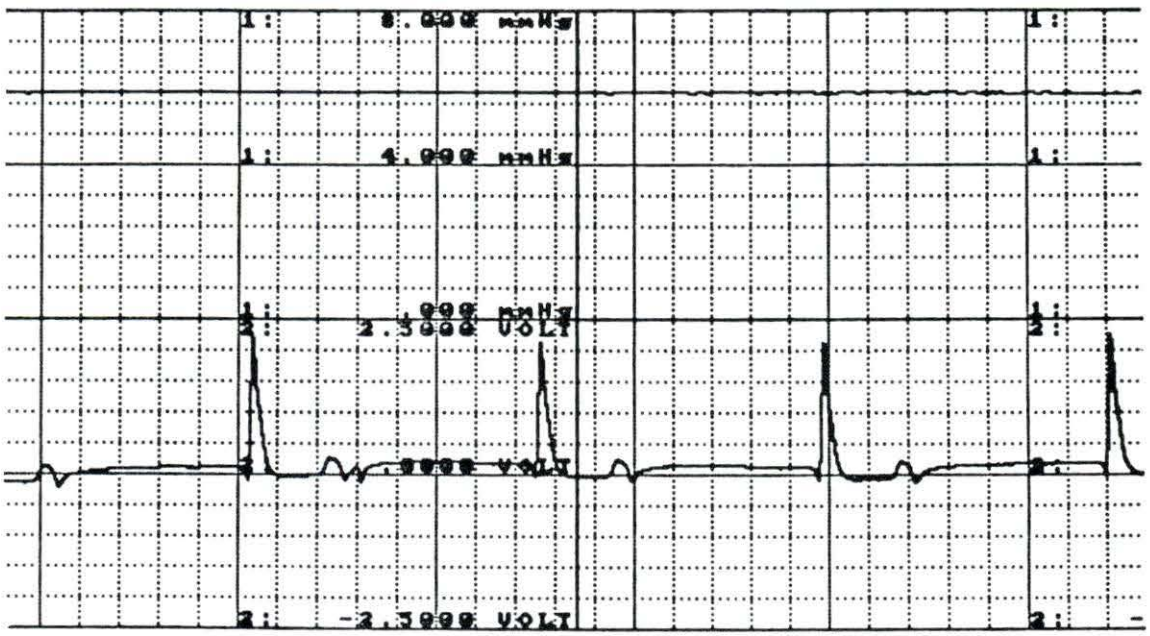

Figure A.7. A recording of venous pressure and the ECG after the heart cuff (bladder empty) was placed on the heart during trial #2

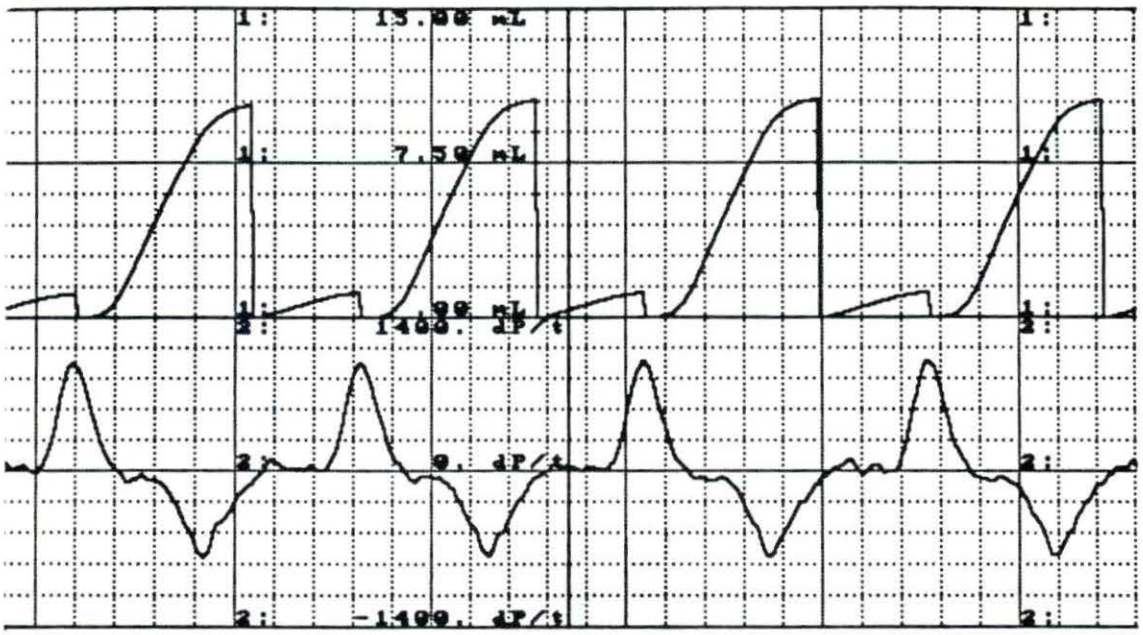

Figure A.8. A recording of stroke volume and contractility after the heart cuff (bladder empty) was placed on the heart during trial #2

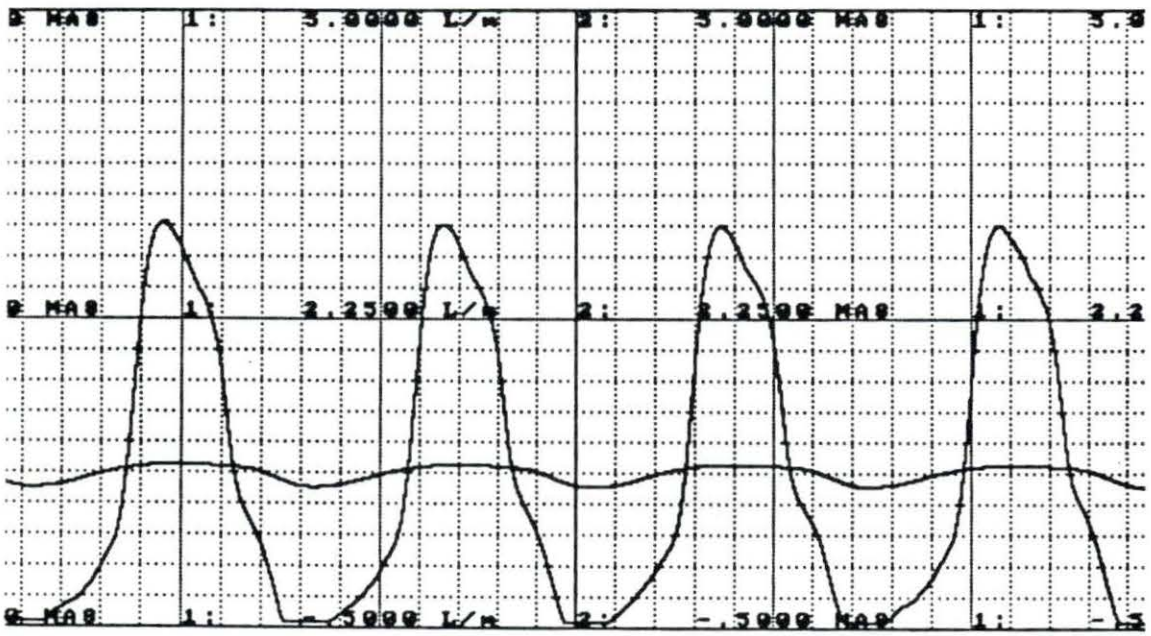

Figure A.9. A recording of aortic flow and mean aortic flow after the bladder was pressurized during trial #2

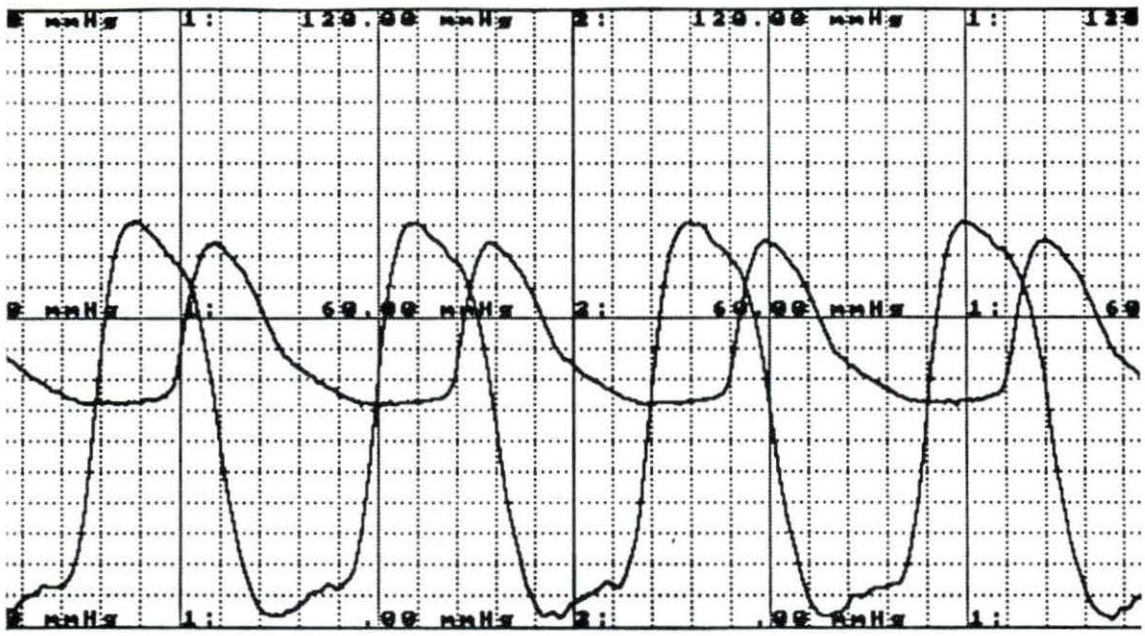

Figure A.10. A recording of aortic and ventricular pressure after the bladder was pressurized during trial #2

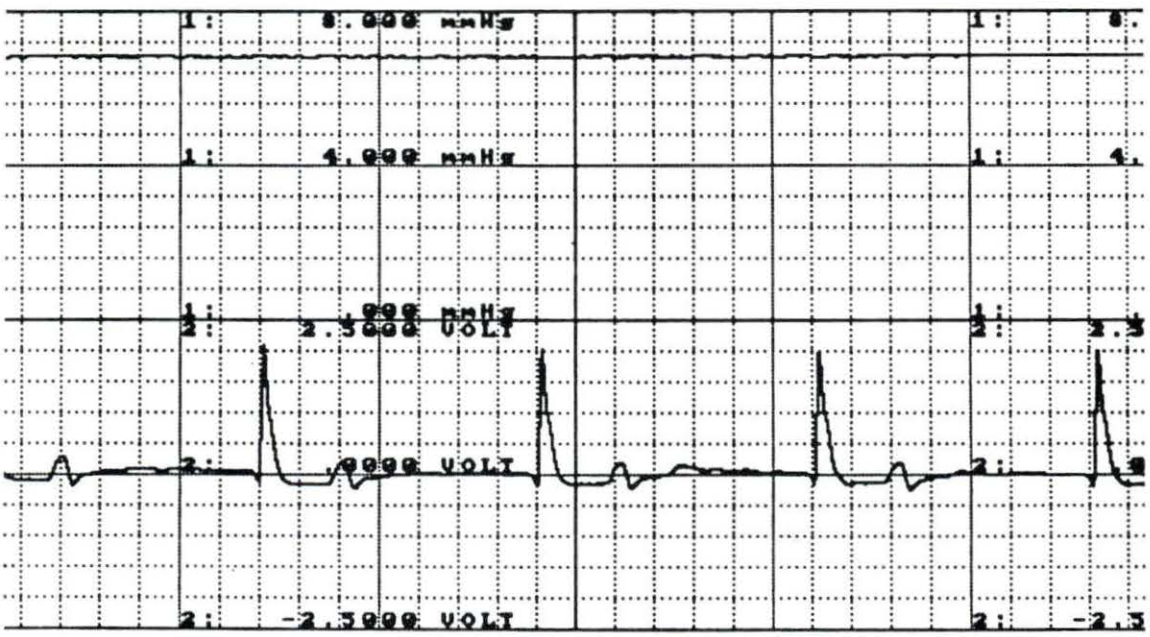

Figure A.11. A recording of venous pressure and the ECG after the bladder was pressurized during trial #2

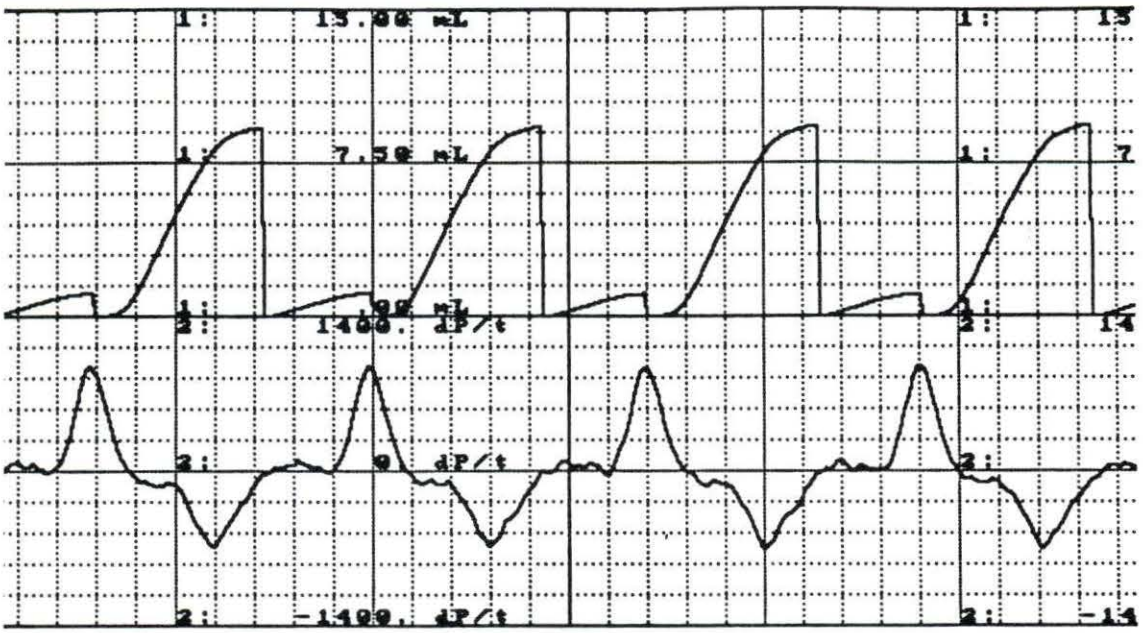

Figure A.12. A recording of stroke volume and contractility after the bladder was pressurized during trial #2

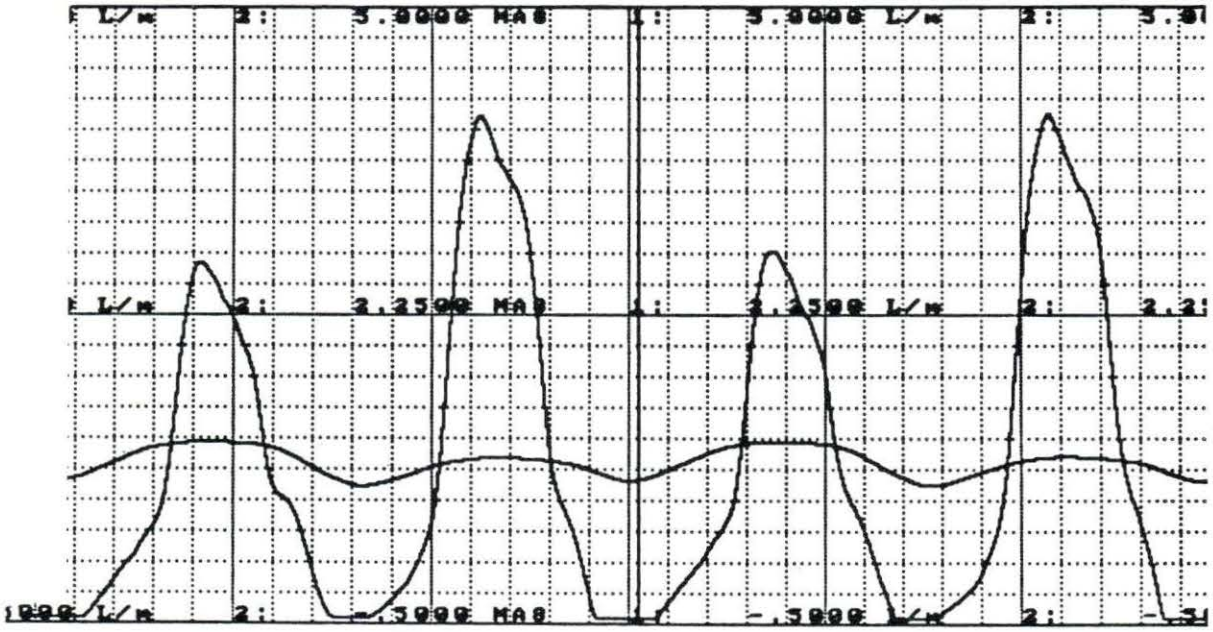

Figure A.13. A recording of aortic flow and mean aortic flow at the start of assisting every other beat during trial #2

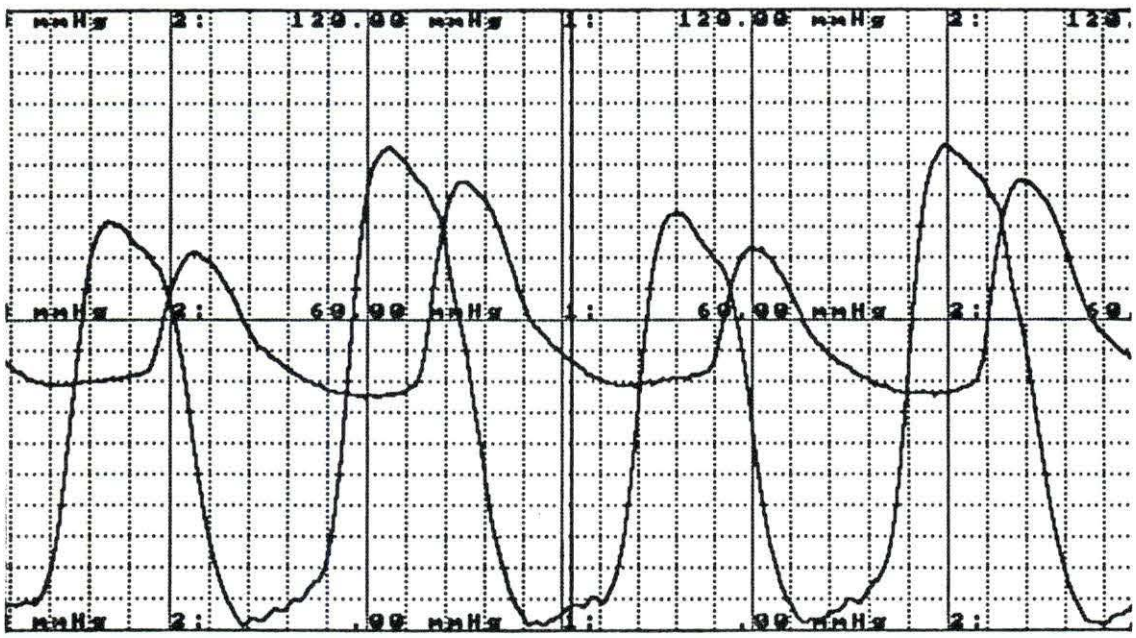

Figure A.14. A recording of aortic and ventricular pressure at the start of assisting every other beat during trial #2

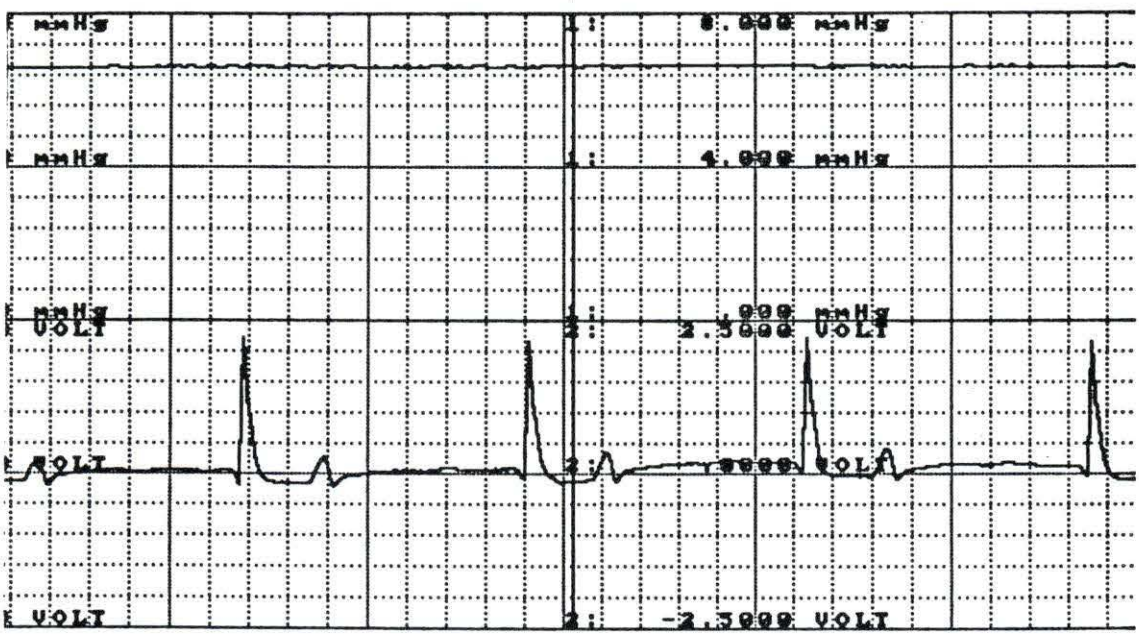

Figure A.15. A recording of venous pressure and the ECG at the start of assisting every other beat during trial #2

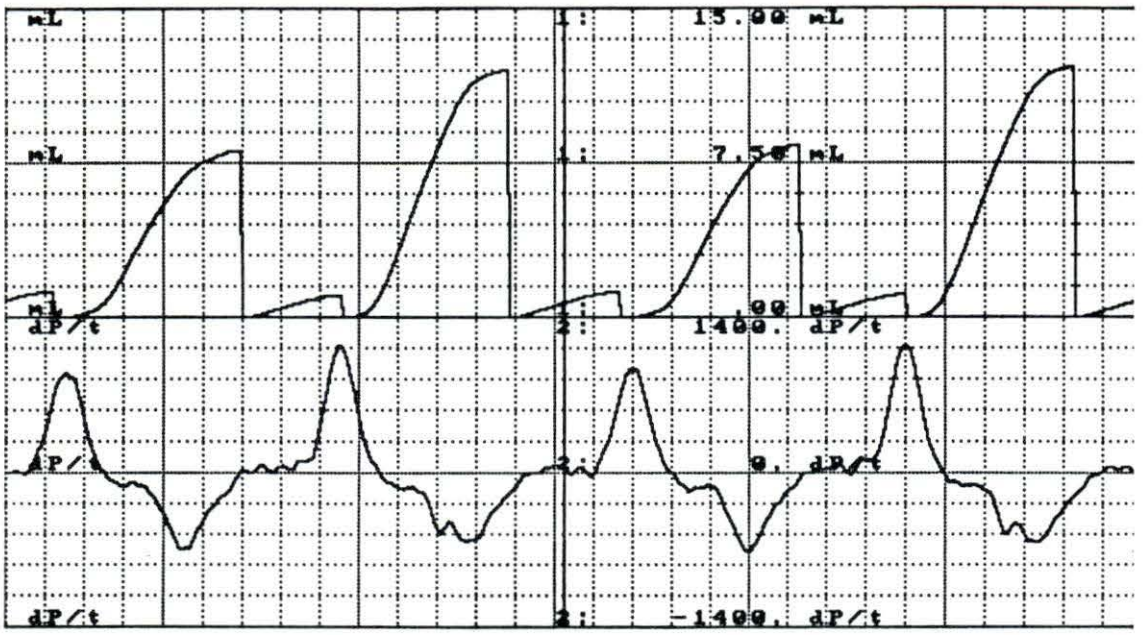

Figure A.16. A recording of stroke volume and contractility at the start of assisting every other beat during trial #2

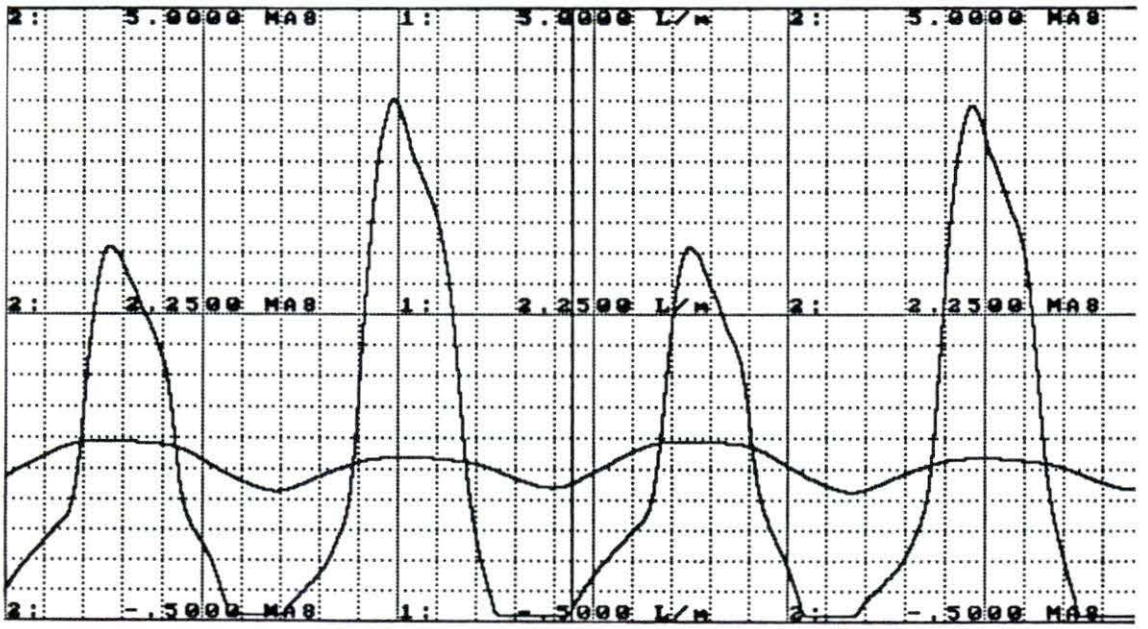

Figure A.17. A recording of aortic flow and mean aortic flow after assisting every other beat for 15 min during trial #2

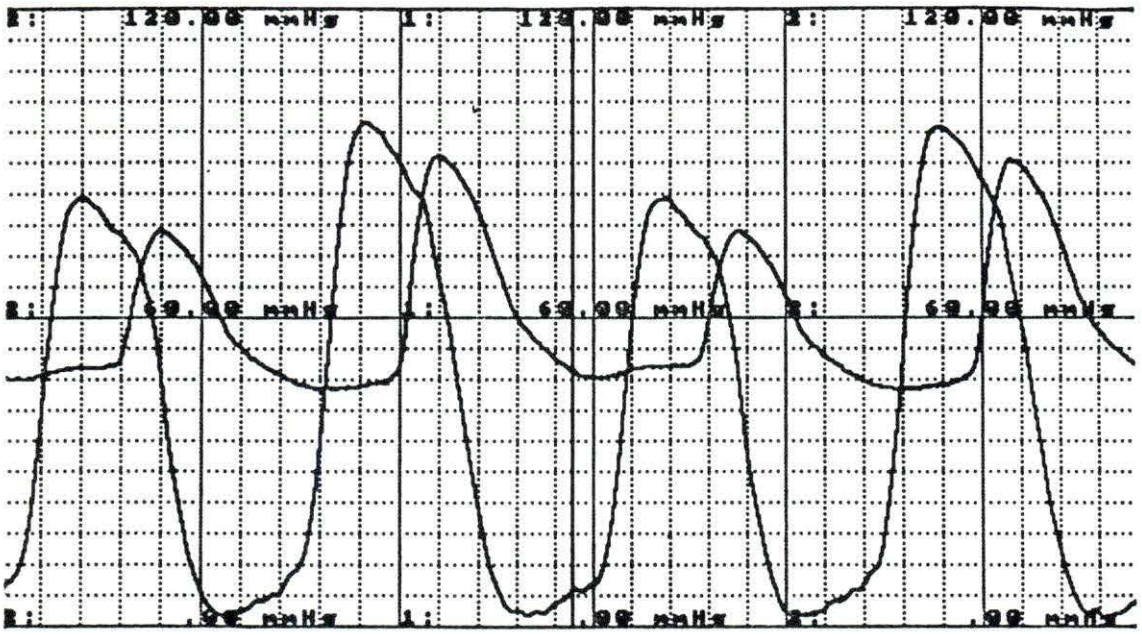

Figure A.18. A recording of aortic and ventricular pressure after assisting every other beat for 15 min during trial #2

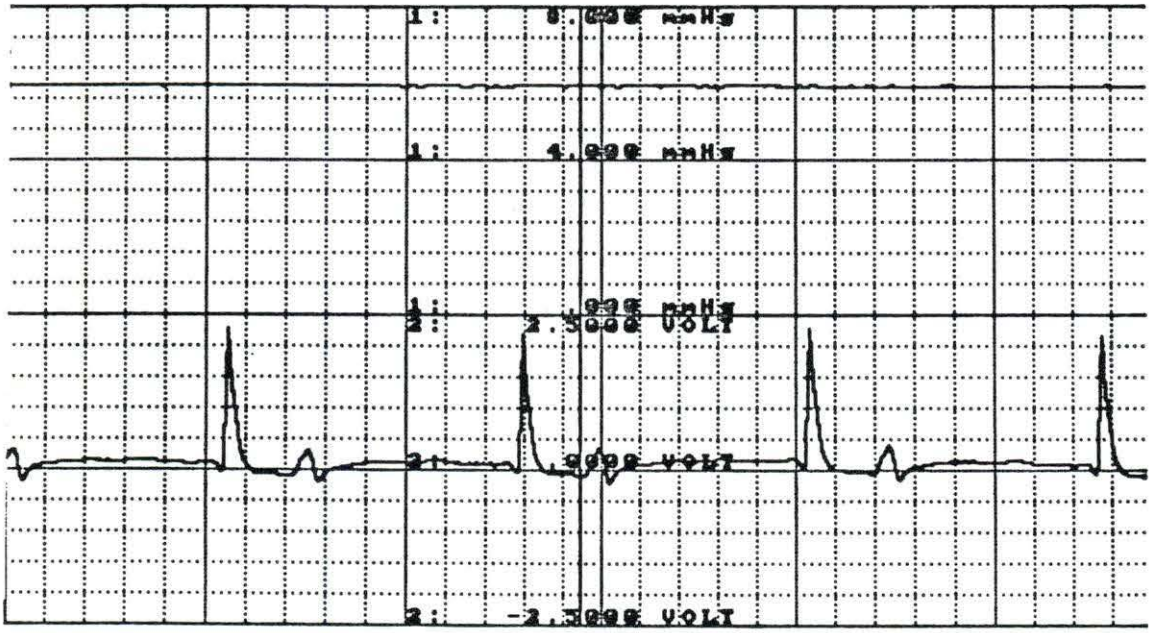

Figure A.19. A recording of venous pressure and the ECG after assisting every other beat for 15 min during trial #2

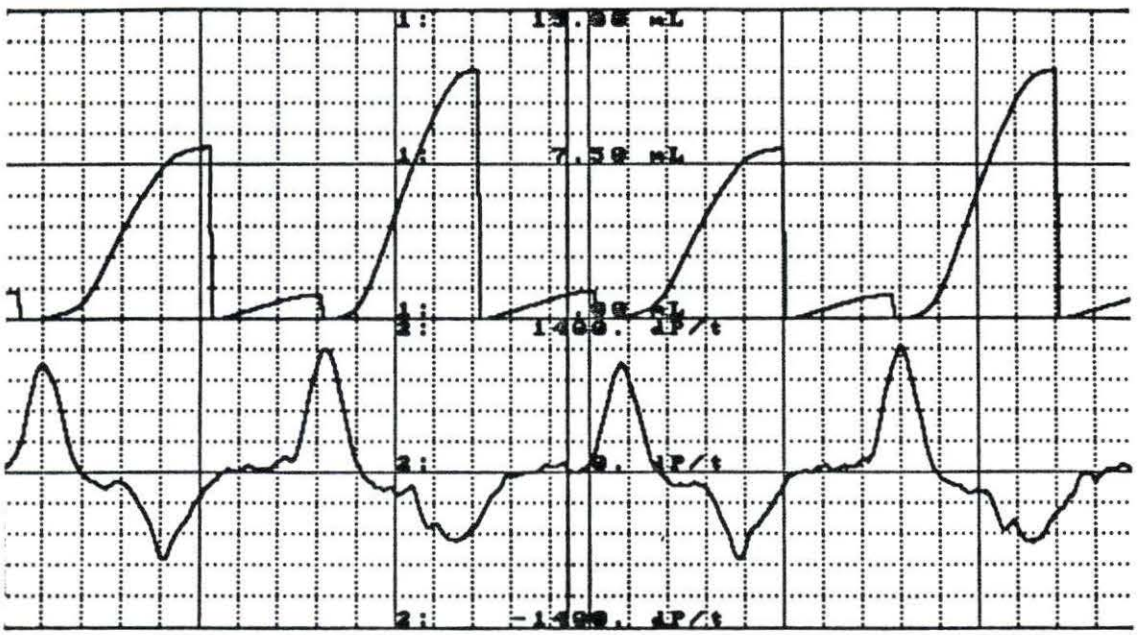

Figure A.20. A recording of stroke volume and contractility after assisting every other beat for 15 min during trial #2

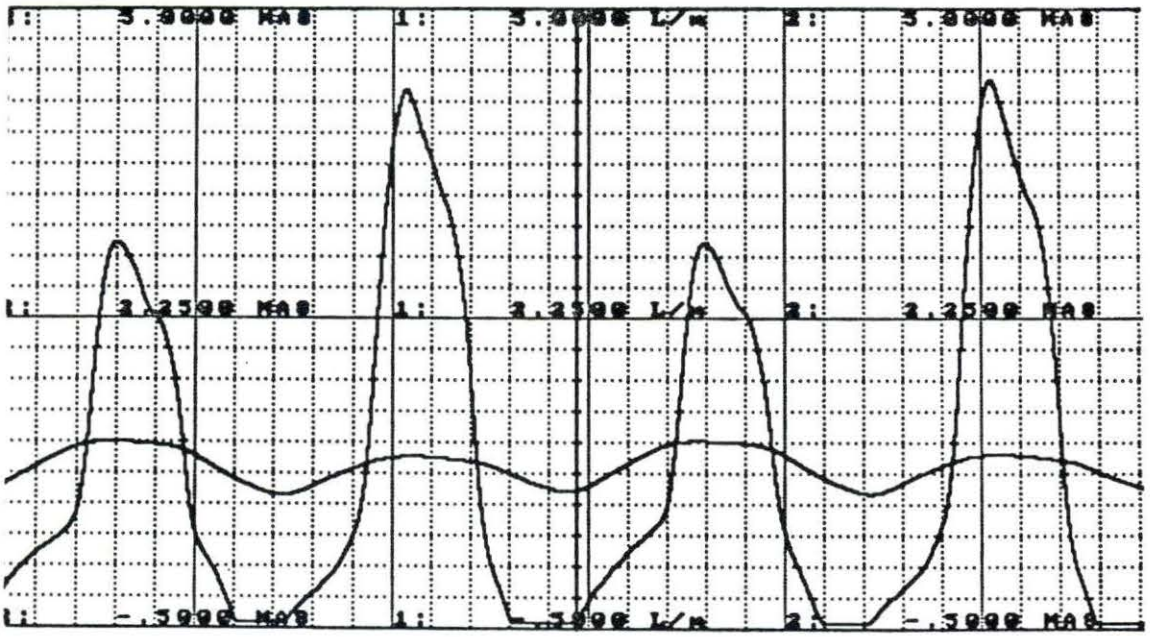

Figure A.21. A recording of aortic flow and mean aortic flow after assisting every other beat for 30 min during trial #2

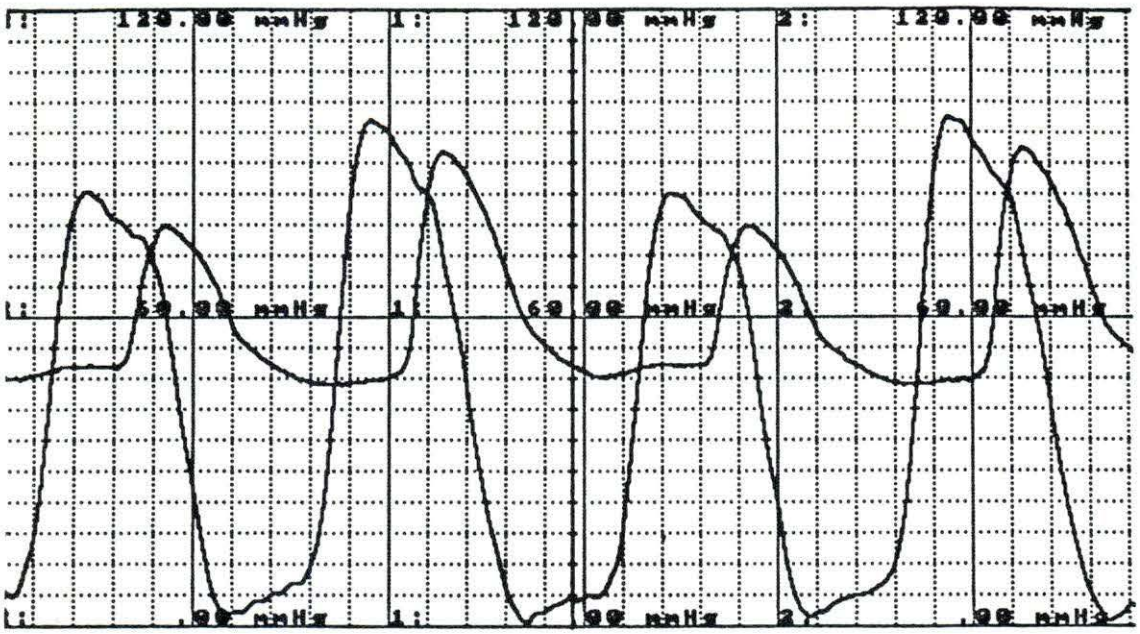

Figure A.22. A recording of aortic and ventricular pressure after assisting every other beat for 30 min during trial #2

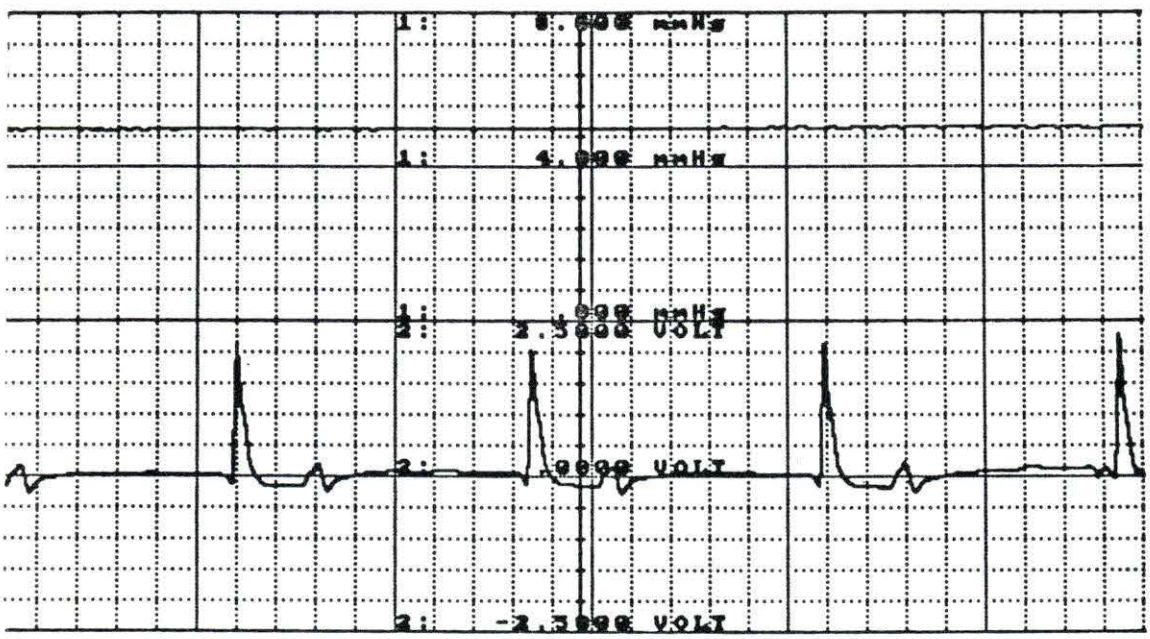

Figure A.23. A recording of venous pressure and the ECG after assisting every other beat for 30 min during trial #2

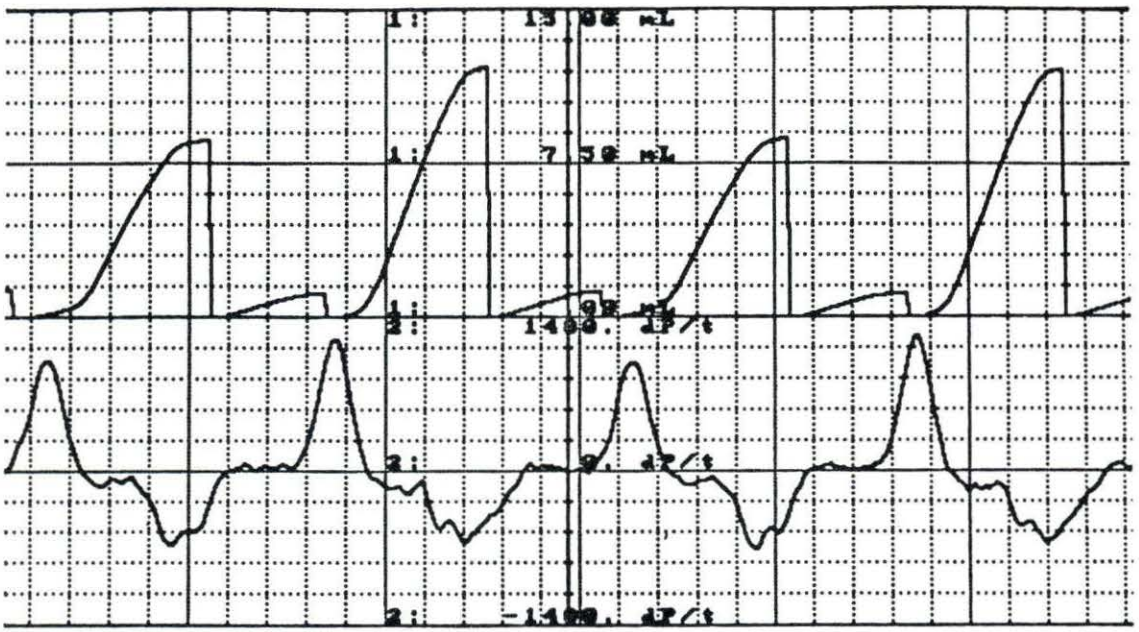

Figure A.24. A recording of stroke volume and contractility after assisting every other beat for 30 min during trial #2

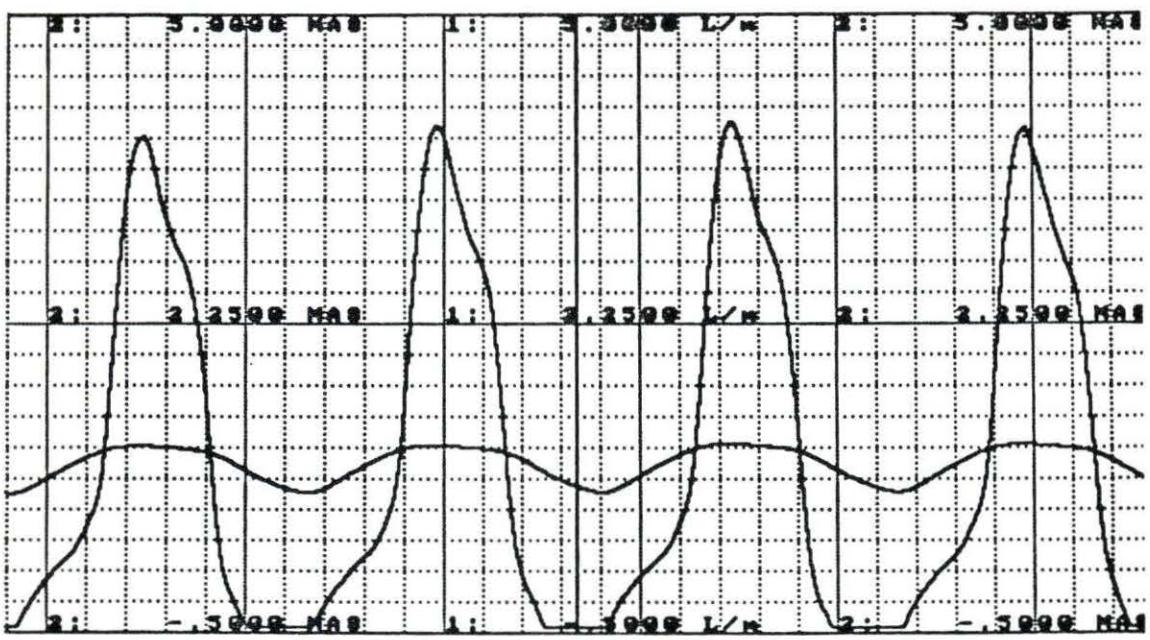

Figure A.25. A recording of aortic flow and mean aortic flow at the start of assisting every beat during trial #2

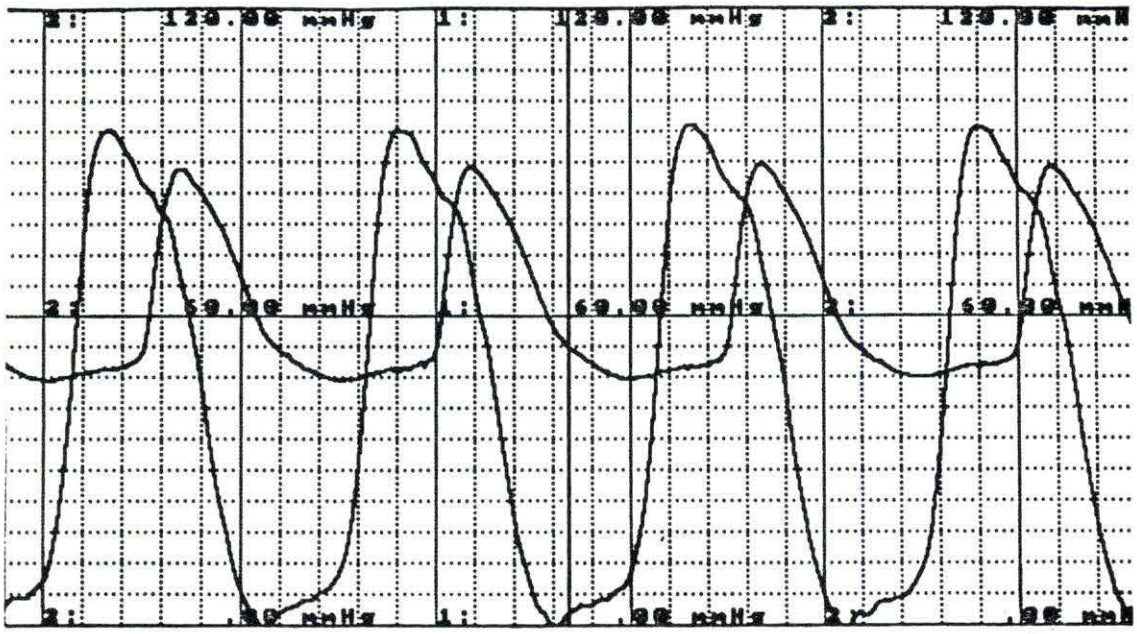

Figure A.26. A recording of aortic and ventricular pressure at the start of assisting every beat during trial #2

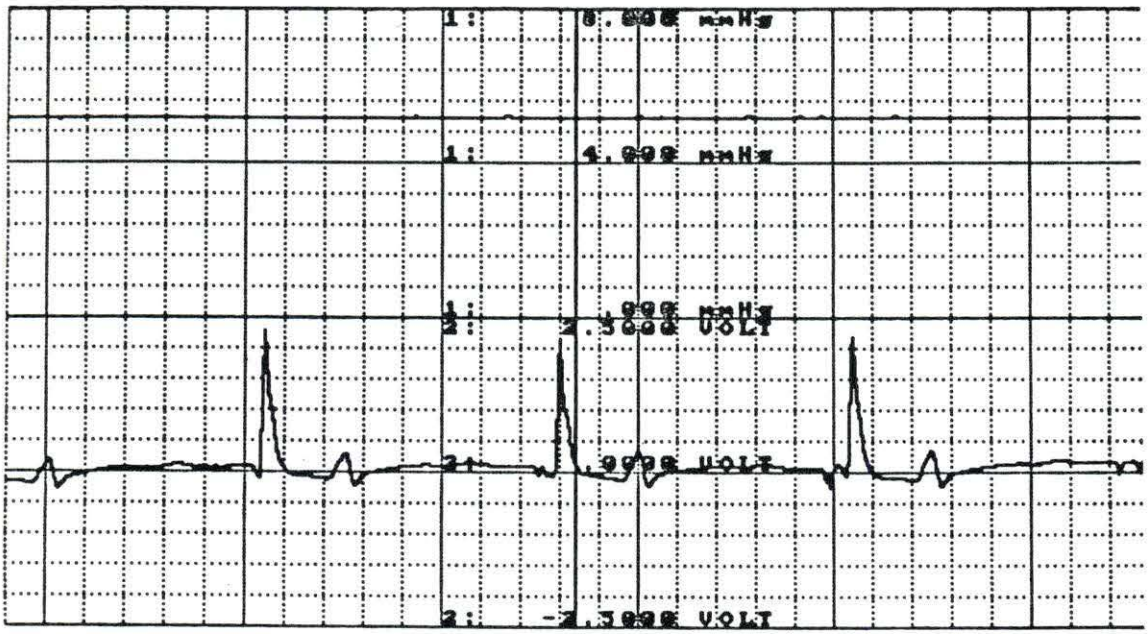

Figure A.27. A recording of venous pressure and the ECG at the start of assisting every beat for during trial #2

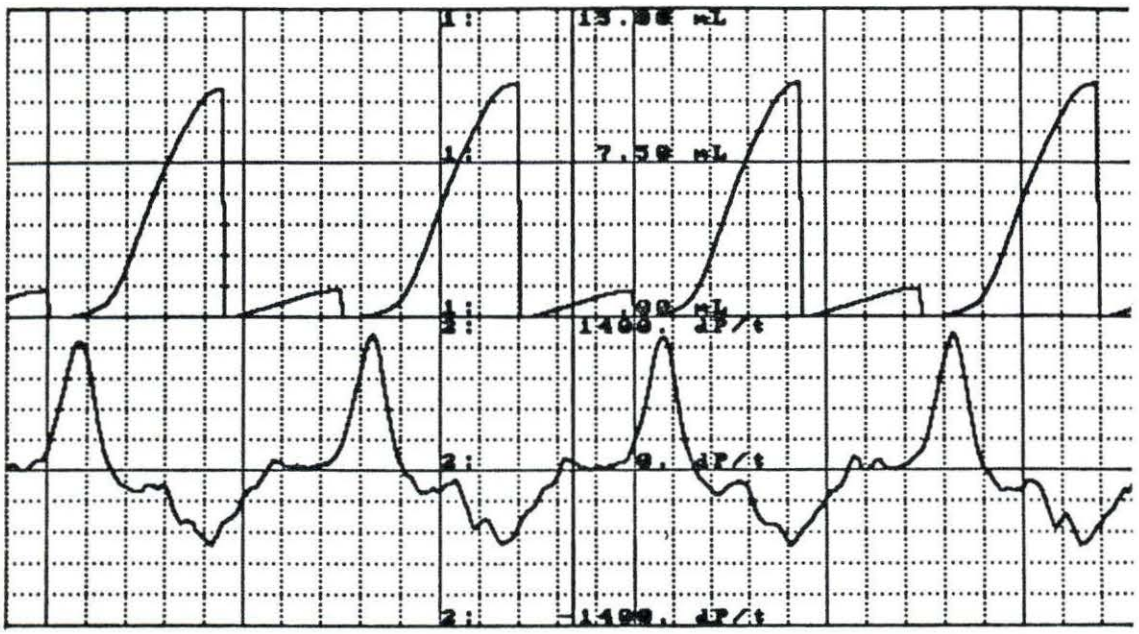

Figure A.28. A recording of stroke volume and contractility at the start of assisting every beat during trial #2

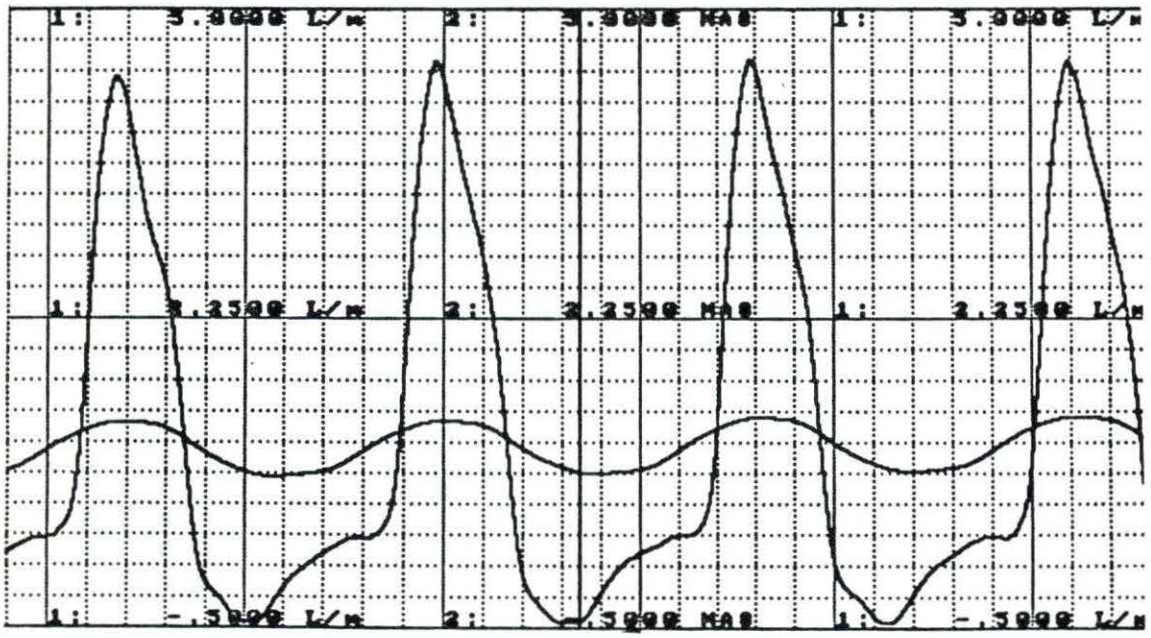

Figure A.29. A recording of aortic flow and mean aortic flow after assisting every beat for 15 min during trial #2

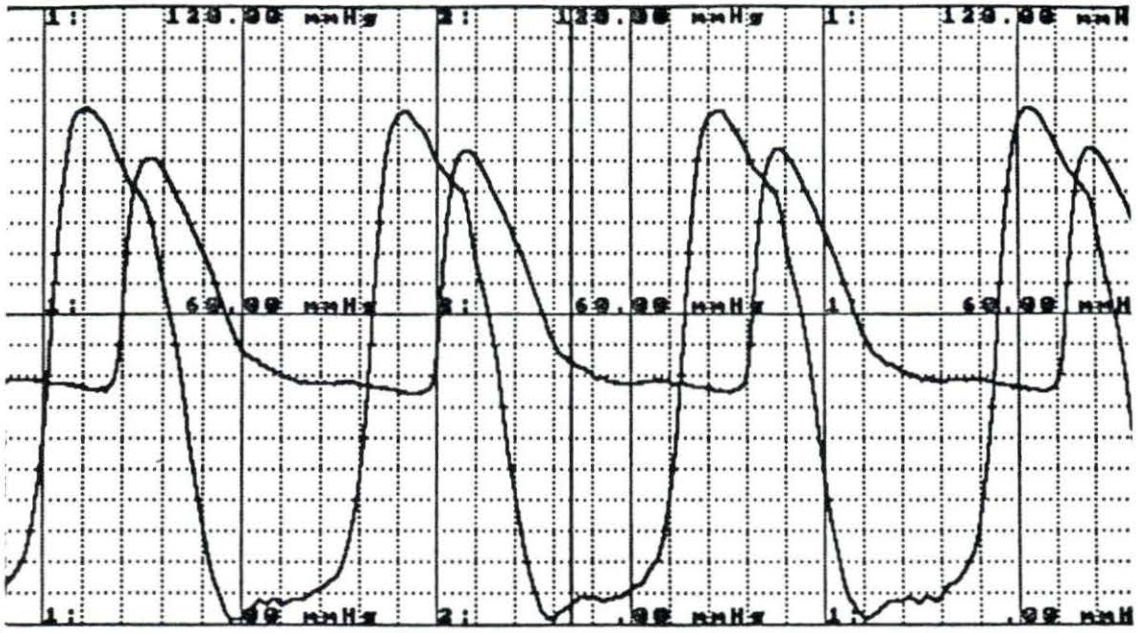

Figure A.30. A recording of aortic and ventricular pressure after assisting every beat for 15 min during trial #2

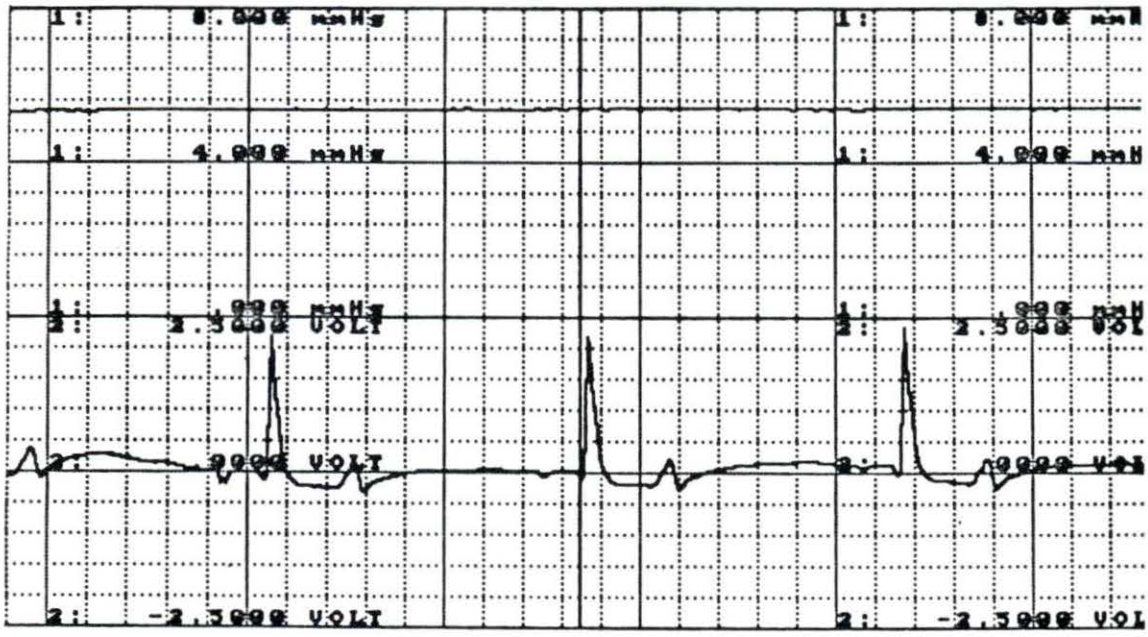

Figure A.31. A recording of venous pressure and the ECG after assisting every beat for 15 min during trial #2

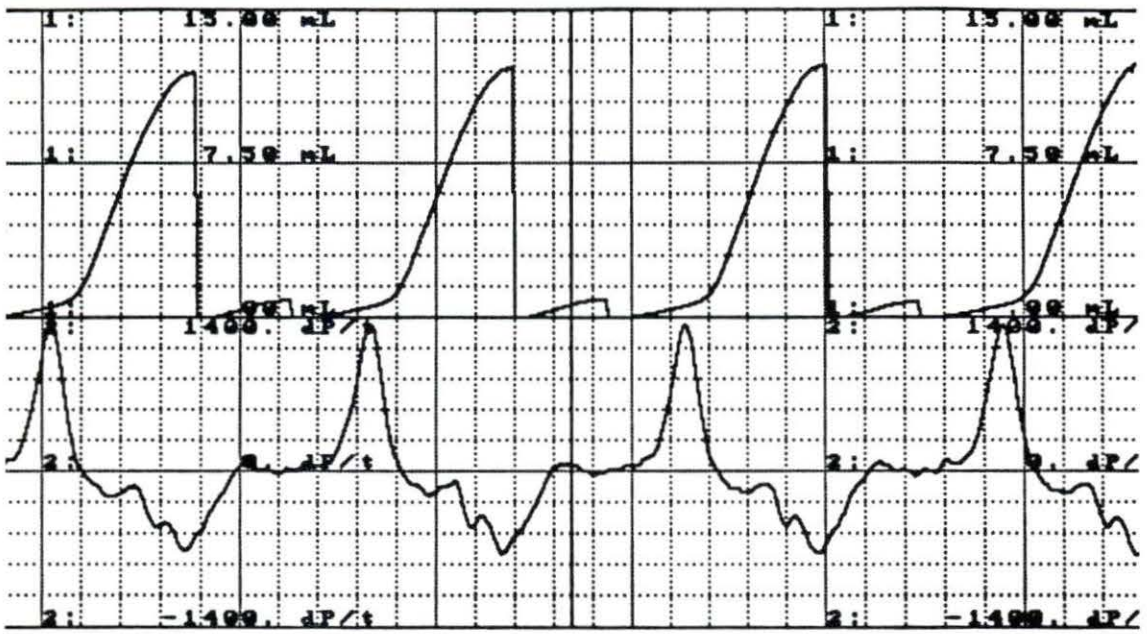

Figure A.32. A recording of stroke volume and contractility after assisting every beat for 15 min during trial #2

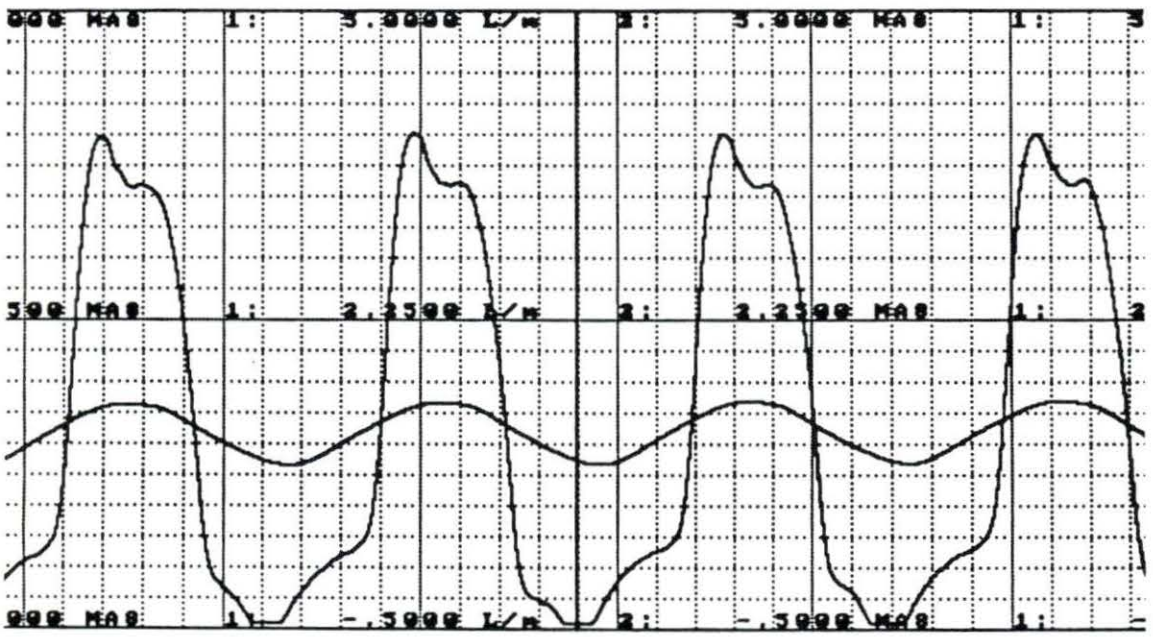

Figure A.33. A recording of aortic flow and mean aortic flow after the heart cuff was removed at the end of trial #2

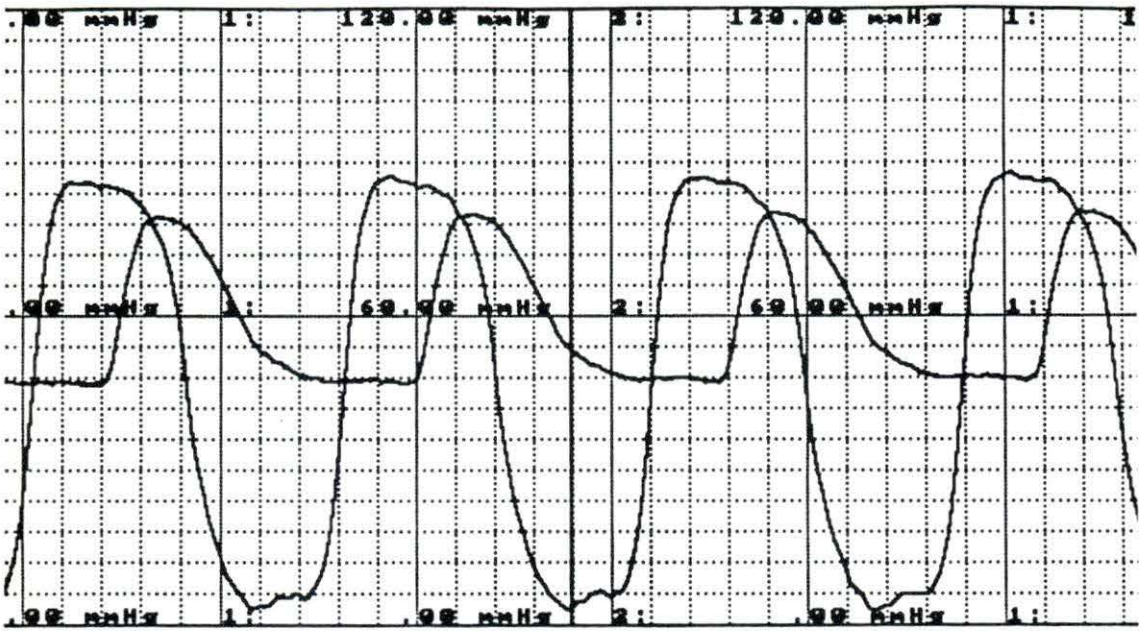

Figure A.34. A recording of aortic and ventricular pressure after the heart cuff was removed at the end trial #2

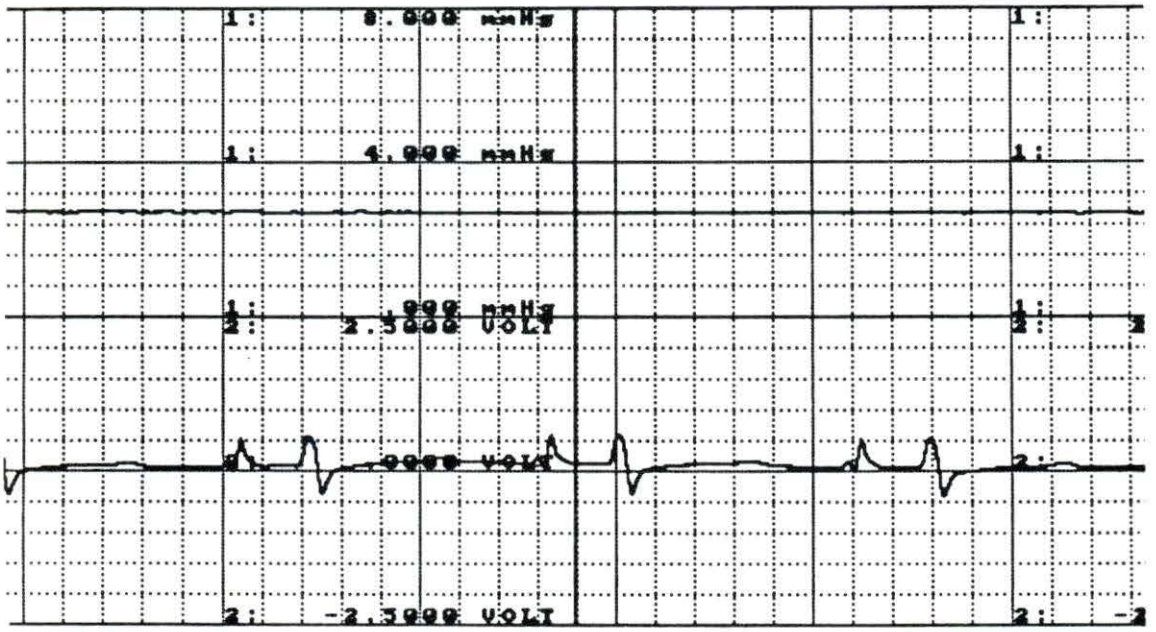

Figure A.35. A recording of central venous pressure and the ECG after the heart cuff was removed at the end of trial #2

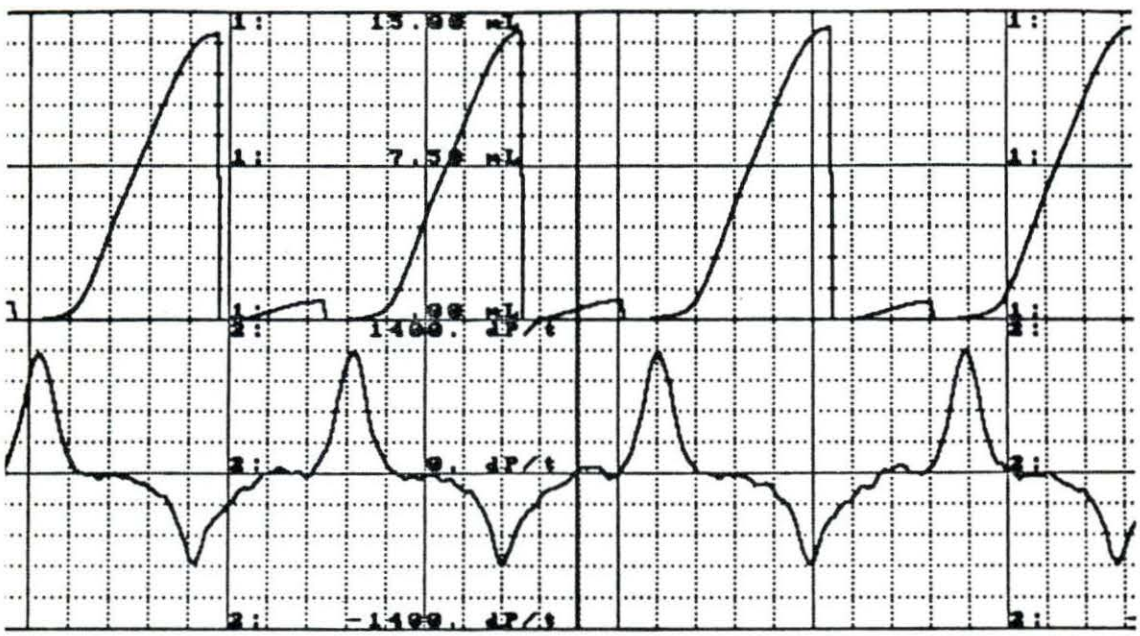

Figure A.36. A recording of stroke volume and contractility after the heart cuff was removed at the end of trial #2