

# ORBITAL FLIGHT PERPARATIONS OF GET AWAY SPECIAL PAYLOAD G-572: DESIGN AND TESTING OF ELECTRICAL SYSTEMS

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## ABSTRACT

Orbital measurements of the cardiac function of Space Shuttle crew members have shown an initial increase in cardiac stroke volume upon entry into weightlessness followed by a gradual reduction in stroke volume. In an effort to explain this response, it was postulated that gravity plays a role in cardiac filling. A mock circulation system was designed to investigate this effect. Preliminary studies carried out on the NASA KC-135 aircraft, which provides brief periods of weightlessness, showed a strong correlation between cardiac filling, stroke volume, and the presence or absence of gravity. The need for extended periods of high quality zero gravity was identified to verify this observation. To accomplish this, the aircraft version of the experiment was reduced in size and fully automated for eventual integration into a Get-Away-Special canister for the orbital version of the experiment. Three nonlinearities, that govern the ability of the apparatus to regulate to a mean cardiac outflow pressure of 95 mm Hg, were identified and minimized. In preparation for an anticipated 1997 shuttle flight, the automated system was again flown aboard the KC-135. The control algorithm was successful in carrying out the experimental protocol, including regulation of mean outflow pressure.

## INTRODUCTION

The advent of the Space Shuttle program provided the initial opportunity to assess cardiac function during space flight. Echocardiographic measurements on crew members documented a reduction in cardiac stroke volume, following an initial increase, after a few days of adaptation to the microgravity environment of space flight<sup>1</sup>. It was postulated that the lack of gravity in orbit plays a role in cardiac filling. On the earth, the orientation of the heart in the upright posture creates a hydrostatic pressure head of about 5 mm Hg. The pressure difference,  $\Delta P$ , is given by the following formula:

$$\Delta P = \rho g \Delta h, \quad (\text{Eq. 1})$$

where:

$\rho$  is the density of the liquid,  
 $g$  is the acceleration due to gravity,  
 $\Delta h$  is the height of the column of liquid.

In the both the right and left ventricles of the heart, the column of liquid is defined as the vertical projection from the base (top) of the heart to the apex (bottom) of the ventricle. In an average adult this dimension is approximately 7 cm with the heart anatomically oriented at 45° to the vertical. In the supine posture, the anterior-posterior  $\Delta h$  is reduced,

but still results in a  $\Delta P$  sufficient to influence cardiac diastolic function. In the "microgravity" environment of orbital flight, hydrostatic pressure is absent because acceleration due to gravity is balanced by centrifugal acceleration. The net acceleration is essentially zero. It was suggested that the absence of this intraventricular hydrostatic pressure difference may be a factor contributing to the observed decreased stroke volume during the period of adaptation to the weightless environment of space flight.

To understand how hydrostatic pressure affects cardiac filling, imagine a simple cardiac model -- a water balloon. As the balloon is filled with water on earth, the transmural pressure of the water pushes out on the sides of the balloon promoting expansion and filling and drawing more water into the balloon. Likewise, as the heart fills with blood, the transmural pressure on the sides and bottom of the cardiac chamber promotes expansion and filling of the heart, drawing more blood into the heart. In space, where hydrostatic pressure is absent, no extra transmural pressure is exerted on the walls of the heart, and therefore less blood enters the cardiac chamber. This counters other factors that results in a net increase in stroke volume and cardiac output prior to adaptation. Since the heart fills passively, the relatively small hydrostatic pressure (about 5 mm Hg) may cause a significant change in diastolic filling. Blood entering

the heart is forced out during systole by the strong, active ejection of the cardiac muscles. Thus when there is a circumstance resulting in reduced cardiac filling, such as the absence of the intraventricular  $\Delta P$ , less blood is pumped per beat.

To explore the hypothesis of acceleration-dependent cardiac filling, a mock circulation system was designed and constructed that is capable of functioning in the absence of gravity. The basic elements included a pumping system (a Utah-100 artificial ventricle<sup>2</sup>), venous and arterial compliance elements, a fixed proximal aortic fluid resistance element and a variable systemic fluid resistance element. The systemic resistance is variable, to allow the adjustment of the mean pressure immediately downstream of the ventricular outflow valve. In adults this pressure, the pressure of the blood exiting the heart into the aorta, is typically 95 mm Hg. As fluid is injected into or withdrawn from the mock circulation system, the atrial pressure will increase or decrease as dictated by the compliance elements. Changes in the atrial pressure directly effect the filling volume of the heart, which relates to the mean fluid flow in the system. This change in flow gives rise to changes in the mean aortic pressure. By adjusting the systemic resistance, the aortic pressure is regulated to 95 mm Hg, but with a new fluid volume in the system. The operating point of the system then represents a different point on the ventricular function curve.

### Ventricular Function Curve

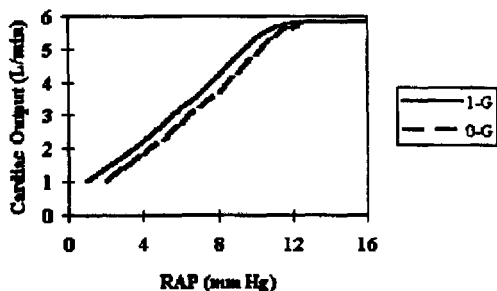


Figure 1 -- The ventricular function curve relates the right atrial pressure to the cardiac output of the heart.

Artificial hearts can be used as a research tool to investigate issues of cardiovascular function. The ventricular function curve (see Figure 1) shows the relationship between the flow of blood pumped by an artificial heart and the atrial pressure (AP), remembering that the AP is the pressure of the blood entering the heart. The steeply sloped part of the curve represents partial filling of the heart during

diastole. Here, a small change in AP results in a large change in cardiac output. In our experiment as the circulating fluid volume of the system is increased, the AP increases, resulting in more blood entering the heart during diastole. Once the AP reaches a level allowing the heart to completely fill, increasing the AP further has little additional effect on filling volume, and thus cardiac output. This is evident by the plateau of the function curve above a certain AP.

Diastolic filling is more properly determined by the pressure difference across the flow resistance presented by the inflow valve. Thus AP and the pressure downstream of the valve in the base of the ventricle (intraventricular base pressure IBP) both influence cardiac output. It is IBP that is directly impacted by gravity. In weightlessness, the pressure inside the ventricle is uniform (neglecting dynamic effects). However, in a gravitational field, hydrostatic pressure, being greatest at the apex, expands the ventricle near the apex. This draws the base outward, also decreasing IBP. This change in IBP is manifested by a shifting to the right of the function curve with decreases in or the total elimination of gravity<sup>3</sup>. The shift occurs since, for the same AP in zero gravity, there is no hydrostatic pressure to promote ventricular filling.

Four developmental versions of the mock circulation system have flown on the KC-135, which provides brief intervals of weightlessness by flying a parabolic flight path. As the plane coasts over the top of the parabola, the cargo and passengers in the aircraft experience about 20 seconds of near-weightlessness. As the plane progresses through the bottom of the parabola it experiences about 40 seconds of a 2-G environment. The results of experiments on this plane showed a strong correlation between hydrostatic pressure and stroke volume. Although the KC-135 flight results were promising, it was evident that, to acquire additional data, extended periods of weightlessness and higher quality weightlessness would be needed. The quality of the weightless environment of the KC-135 is low because of mechanical and aerodynamic noise effects of the plane flying through the air. Rather than a steady weightless environment there is an appreciable amount of vibrational noise. The brief duration of the weightlessness on the aircraft did not allow sufficient time to servo-regulate the systemic resistance element to the altered hemodynamic state.

In order to obtain high quality, extended microgravity data during orbital flight, several refinements needed to be made to the initial design of the mock circulation system, which consisted of a large instrument chassis and an experimental platform

which weighed over 600 pounds. A redesign was required for integration into a Get-Away-Special (GAS) canister for an orbital experiment which limits the weight of the payload to 200 pounds and the volume to 5.0 cubic feet. This redesign of the experiment included the conversion from manual adjustments to a fully automated, electronically controlled system. Since the GAS canisters are completely self contained in the cargo bay of the Space Shuttle, the electronic controller must control all equipment necessary to run the experiment in the GAS can, supervise the protocol of the experiment, and initiate data collection.

**METHODS**

The mock circulation system developed for orbital flight is a fluid circuit consisting of seven elements (see Figure 2); 1) a pump, 2) a fixed

resistance element, 3) an arterial compliance chamber, 4) a variable systemic fluid resistance element, 5) a venous compliance chamber, 6) an artificial atrium, and 7) a fluid volume reservoir and infusion/withdrawal pump. The pump is the left ventricle of a Utah-100 total artificial heart. The compliance chambers (modified Penn State<sup>4</sup>) emulate the compliance, or elasticity, of the arteries and veins. The atrium (modified Penn State<sup>4</sup>) is a thin spherical chamber directly outside of the inflow valve (mitral valve) of the heart, that acts like a reservoir. The fluid resistor consists of a porous plastic sheet mounted in the fluid path and a resistor plate (see Figure 3). The resistor plate can be adjusted up and down to expose varying areas of the resistive material. With this configuration the conductance is directly proportional to the exposed area of the resistive material. The servo-regulation of the resistor plate

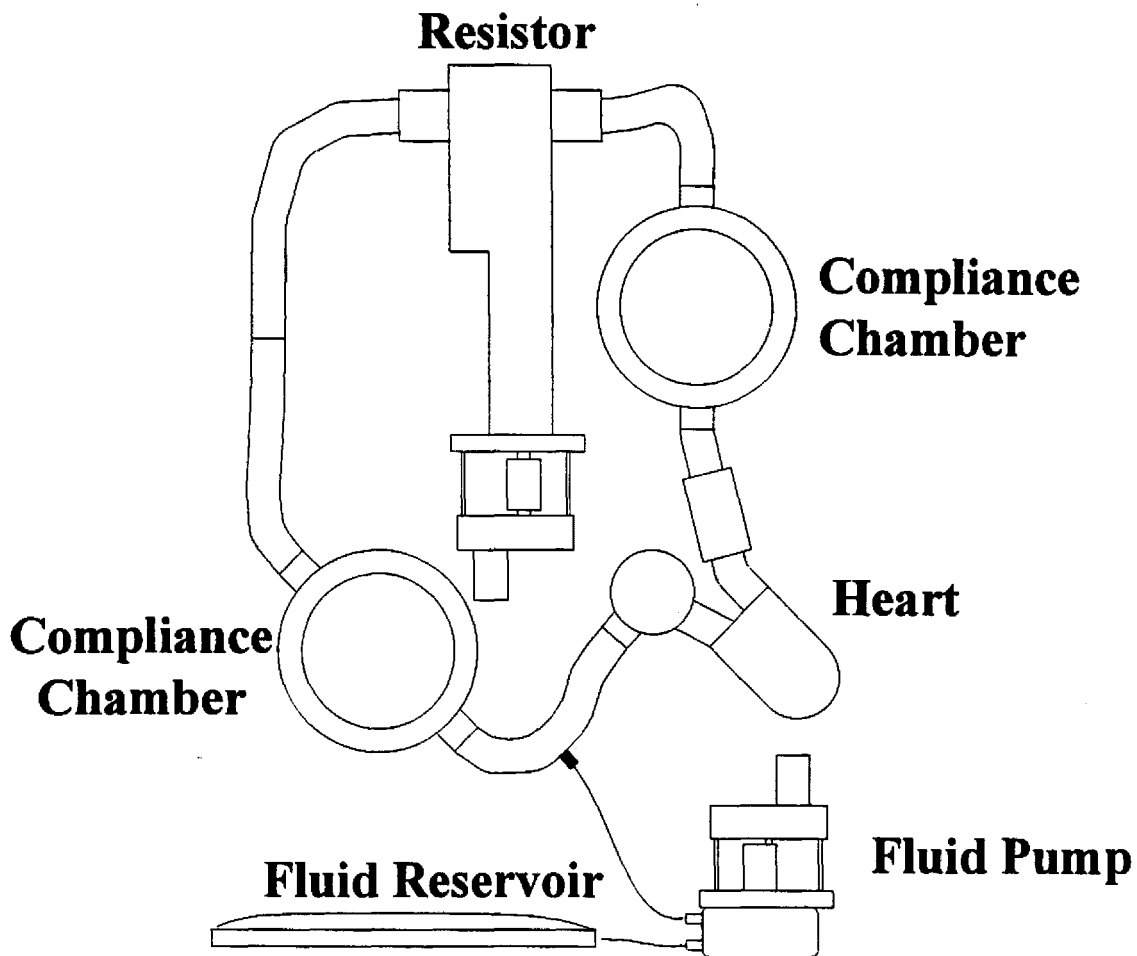


Figure 2 -- The experimental layout including all seven system components.

occurs following specified events in the experimental protocol.

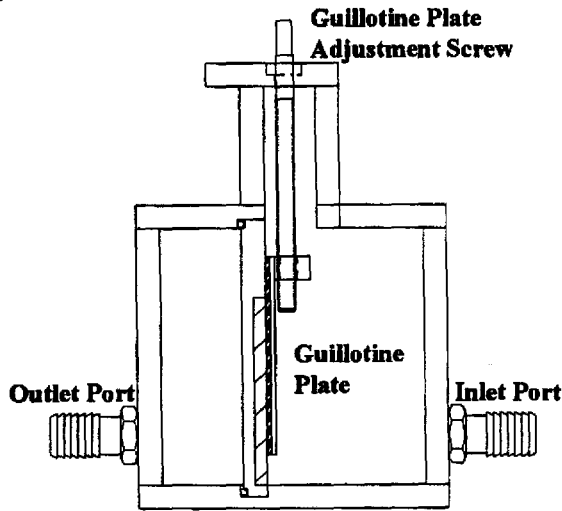


Figure 3 -- Shows the internal workings of the variable fluid resistive element.

The steps in the protocol of the orbital experiment are explained below and are shown pictorially in Figure 4. Details of the protocol are given in the sections that follow.

Once the experiment is activated after launch by shuttle astronauts, the temperature is measured to determine if heating is required. If needed, the fluid is heated until one of the following conditions is met a) the desired temperature is achieved, b) three hours have passed, or c) the heater batteries are depleted. After heating, the instrumentation is powered. This includes the ultrasonic flow meter board (Transonic Systems), catheter tip pressure transducers (Millar Instruments), seven channel data recorder (TEAC), accelerometer (Endevco), and the Heimes Heart Driver™ (Cardiwest Technologies). When the experiment is initiated, the resistor plate is fully opened and fluid is injected or withdrawn to obtain a mean outflow pressure of 95 mm Hg, at which time data is collected. As a calibration step, the heart driver is then turned off and again data is collected. Next the heart driver is turned back on and a predetermined quantity of fluid is withdrawn to move the system to a new point on the cardiac function curve. The resistor plate is then adjusted to obtain a mean outflow pressure of 95 mm Hg and data is collected. This cycle continues, approaching the bottom of the ventricular function curve, until the resistor plate is fully closed.

After the resistor plate is fully closed, the series of operating points is repeated in the opposite

order by incrementally injecting fluid. For each point, the resistor plate is adjusted to obtain a mean outflow pressure of 95 mm Hg. Data is collected first with the heart driver on and then with it off. At the end of the second series of operating points, the pressure probes are vented to the canister ambient pressure for a zero pressure calibration measurement. At this point, the set of experimental data is complete, however to increase the number of samples, the experimental protocol just described is repeated until the batteries are depleted or the astronauts terminate the experiment (about three hours after experiment initiation).

A key process to the proper function of the experiment is the ability to adjust the mean outflow pressure by adjusting the exposed area of the porous resistive material by moving the resistor plate. This plate is positioned via a 12 volt DC motor, connected to the resistor plate by a drive screw. The motor is wired in a standard H-bridge configuration which allows bi-directional operation from a single power supply. The H-bridge is composed of four double-pole double-throw relays. It was not necessary to use either the double-pole or the double-throw options, however, they were chosen to provide redundancy in the design. Logic circuitry is used to insure that relays cannot be opened in a way that shorts the power supply.

The position of the resistor plate is sensed only at the extremes (fully open and fully closed) to limit the travel of the plate. Two inductive position sensors are employed to sense the presence of a block of metal mounted on the resistor plate. The sensors operate from a 15 volt power source. They pass a low current when they are triggered, and a higher current when they are not triggered. Each sensor is used in conjunction with a voltage divider circuit to provide a signal that can be converted into the TTL voltage standard. The conversion is made by applying the signal from the voltage divider to a 74LS04 inverter chip (Motorola). The output signal is read by the microcontroller, which operates the motors appropriately.

The pressure in the mock circulation system is sensed at five locations. The pressures are sensed using high fidelity catheter tipped pressure transducers (Millar Instruments). Of the five pressures that are recorded, only the outflow or aortic pressure (see Figure 4) is used to position the resistor plate.

In addition to the five main elements in the fluid circuit, there is also an infusion-withdrawal pump for changing the circulating fluid volume of the system. When fluid is injected into the system or

## Experimental Protocol

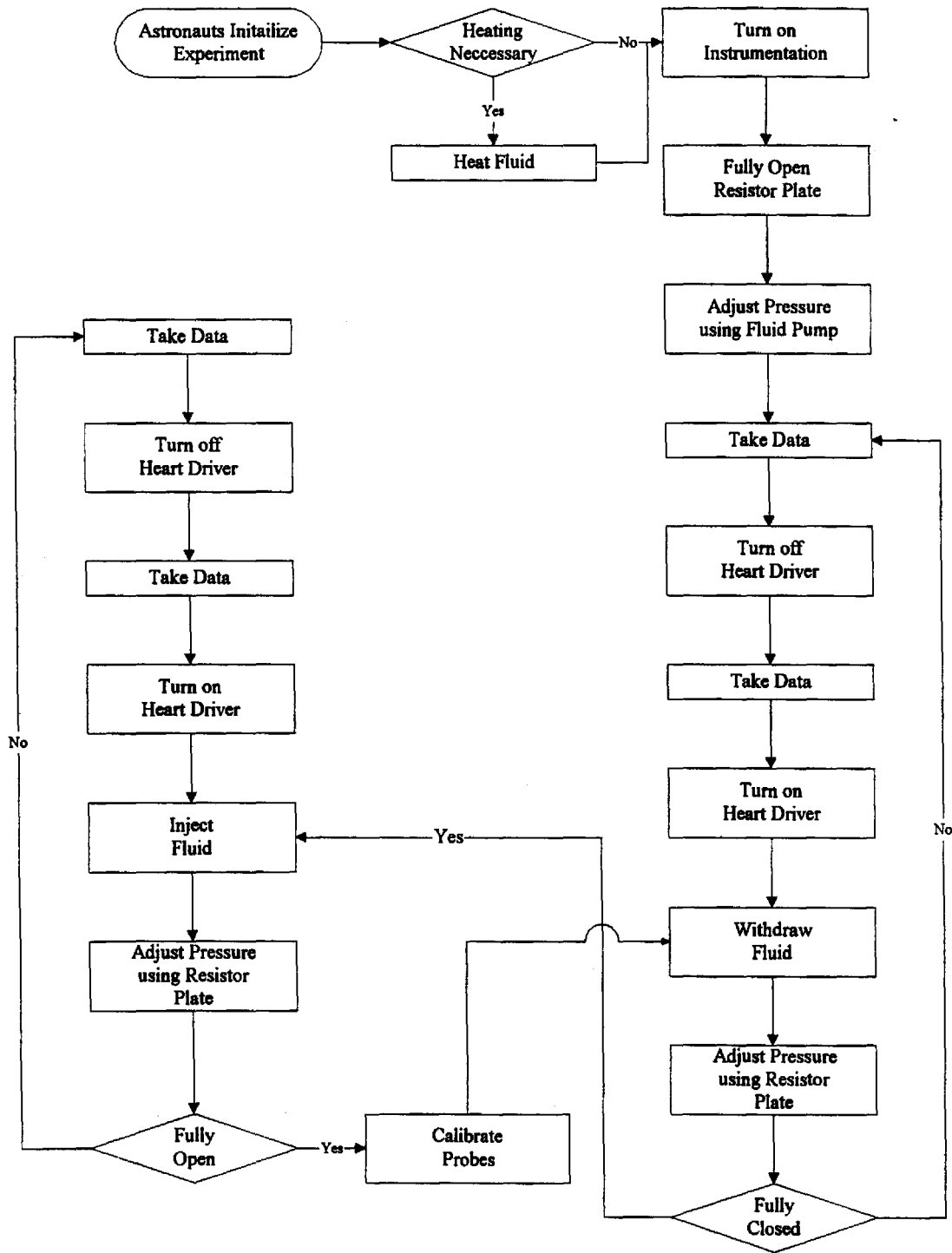


Figure 4 -- A flow chart illustrating the protocol of the orbital GAS canister experiment

withdrawn from the system, the mean circulatory filling pressure of the system changes. The goal of the controller is to bring the mean aortic pressure back to a preset pressure condition, in this case 95 mm Hg (see Figure 5). The only parameters that are known to the controller are the present mean pressure and the desired mean pressure.

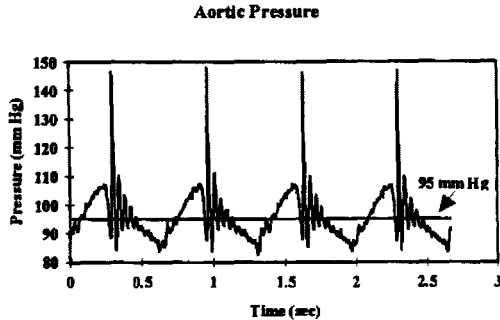


Figure 5 -- The aortic pressure as a function of time. The horizontal line represents the target mean pressure of 95 mm Hg.

There are several ways that pressure regulation can be obtained. Ideally we would like to know the relationship between the position of the resistor plate and the pressure, or at least the relationship between the change in position and the change in pressure. In the mock circulation system there is a second parameter associated with mean pressure, the circulating fluid volume of the system. Unfortunately, for each volume there is a different function,  $\Delta p_k = f(\Delta x_k)$ , where  $k$  is the increment of fluid volume. This function relates the change in resistor plate position,  $\Delta x_k$ , to a change in pressure,  $\Delta p_k$ . It is impracticable to keep track of the exact fluid volume, because of imprecision in the infusion-withdrawal pump and the possibility of fluid exiting the system. There is also no accurate way to determine the exact volume of the system once the experiment is in progress without draining the fluid. This being the case, we can not determine the parameter  $k$  and, therefore, cannot make use of the function  $f(\Delta x_k)$ . Further, this function is not stationary, it changes over time as parameters of the fluid resistance change. These changes occur because of the filtering action of the resistor. As the resistor filters the fluid, its resistance increases, changing the entire family of functions,  $\Delta p_k = f(\Delta x_k)$ . It should also be mentioned that the resistance is also a function of fluid viscosity, which is a function of temperature. This is why a provision is made to heat the circulating fluid in the experimental protocol.

Having considered the requirements of this dynamic system we chose a straight forward control scheme. Since data to construct a ventricular function curve is derived from values obtained only after a steady state condition has been reached, when the mean pressure is constant with respect to time, we are not constrained to a maximum settling time. That is, we have a relatively unconstrained amount of time to adjust the mean pressure back to 95 mm Hg after the injection or withdrawal of fluid. The pressure is measured and the adjustment motor is powered for a time  $\Delta t_1$ , in a way that minimizes:

$$|\Delta p| = |\bar{p}_{\text{current}} - \bar{p}_{\text{desired}}|.$$

The time  $\Delta t_1$  is empirically chosen to be small enough so that the resistor plate does not cross the target pressure window in one interval, that is to say, no overshoot is allowed. The mean pressure is measured again, and the cycle repeated, until the mean pressure is within the desired pressure window.

## RESULTS

In preliminary testing, we defined the target pressure window, centered around a pressure of 95 mm Hg, to be  $\pm 2$  mm Hg. With system and algorithm refinement, the goal is to reduce this window to  $\pm 0.5$  mm Hg. To insure the mean pressure has entered the window, the system is forced to obtain an acceptable reading three consecutive times. The system has taken, on average, three movements of the resistor plate to return the average pressure to the target zone.

The lower graph in figure 6 shows the regulation process. As part of experimental protocol fluid is either injected into or withdrawn from the system. This example assumes that fluid is being withdrawn from the system. Step one is immediately after the heart driver has been turned on. During this step fluid is withdrawn and the average pressure drops. Step two shows where regulation begins. The withdrawal of fluid has dropped the average pressure by 5 mm Hg. The computer samples the instantaneous pressure signal and integrates for a predetermined period of time to obtain the average pressure. It then decides on a course of action. In this case it closes the resistor plate raising the average pressure slightly and then it resamples the pressure. This continues until it has sampled the pressure three consecutive times and found the mean pressure to be within the desired pressure tolerance. The computer then shuts off the heart driver briefly as a calibration step and the cycle repeats. The full cycle can be seen clearly in the top

graph of figure 5 which shows the on and off cycles of the heart driver. The instantaneous pressure is shown in the middle of the average pressure to emphasize that averaging is a postsampling process.

It was previously stated that in steady state the mean pressure does not change with respect to time, but this is not strictly true. Steady state means the change in volume with respect to time is zero,  $dV/dt=0$ , and that all the changes in mean pressure caused by any changes in volume have died out. If the

mean was truly constant at steady state then the pressure would be a stationary ergodic process<sup>4</sup>, stationary<sup>5</sup> in the mean. It is not. The mean randomly changes from heart beat to heart beat, having a mean equal to 95 mm Hg and an undetermined standard deviation. This resulting error can be made arbitrarily small by integrating for a sufficient length of time.

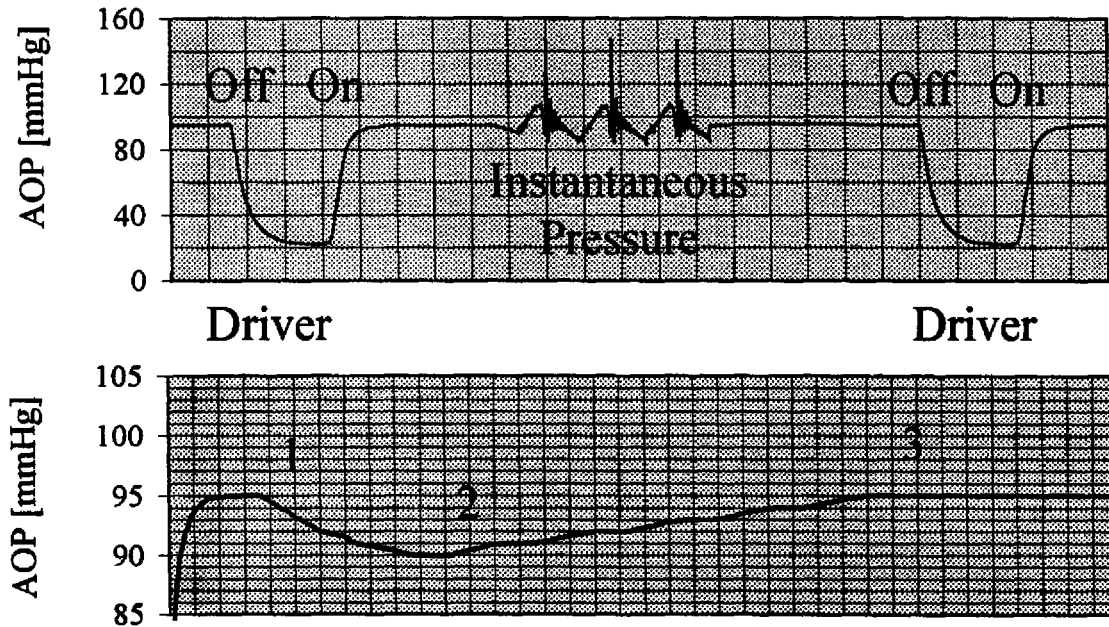


Figure 6 -- TOP: Shows one experimental cycle, which starts when heart driver is turned on and continues until it is turned off. The mean pressure is shown with the instantaneous pressure displayed mid-cycle. BOTTOM: 1) Fluid is withdrawn from the system, 2) regulation begins -- resistor plate is incrementally closed, 3) mean pressure is 95 mm Hg -- system is ready for data collection with heart driver on.

In the development of the current mock circulation system, three major nonlinearities were uncovered. They were large enough to affect the local stability of the system. These nonlinearities manifested themselves as intermittent oscillations as the system was trying to settle at 95 mm Hg. At times the system would not settle and had to be reset. The first step in correcting these problems was to characterize the oscillations and record the conditions present when oscillations began.

The first nonlinearity was more of a measurement error than a nonlinearity. The pressure wave form has two major components; a constant pressure offset (the mean), and a periodic component. The problem arose because the period of the oscillatory component of the heart rate was not precisely known, in fact it is not constant but varies

slightly with each heart beat. In calculating the average<sup>6</sup>, the pressure wave form was integrated for time  $t$  and then normalized by dividing by  $t$ . The imprecision in  $T$ , the period, makes it difficult to choose  $t = nT$  and gives rise to the imprecision in  $P_{average}$ . This error can be decreased by integrating for a longer time, after assuming a reasonable value for  $T$ . The orbital experiment will be allotted a fixed time to complete its task. By effectively increasing the settling time, the amount of data that can be collected is decreased, an undesirable consequence. An integration time of 4.5 seconds was chosen to give the best balance between precision and speed.

The second nonlinearity caused the average aortic pressure to jump by 20 to 40 mm Hg in one heart beat and at apparently random times. Once in this new state, it was stable and would remain there

for some time. The source of this fluctuation was traced to the artificial atrium. The atrium is constructed of a very thin compliant material (Biomer, Ethicon). It is spherical in shape for positive transmural pressures and collapses for large negative transmural pressures, and was not expected to affect the aortic pressure significantly. However, it was determined that the spherical structure of the atrium allowed two stable shapes for a range of small negative atrial pressures, one when the atrium was completely full to distension and another when the atrium had partially collapsed. As the resistor plate was opened, acting to decrease aortic pressure, the aortic pressure would decrease slowly and then suddenly drop. When the peak atrial pressure reached a critical point, the atrium would fully inflate and then not recollapse as the instantaneous pressure decreased. The observed result was a sudden drop in average aortic pressure. In the other direction (closing the resistor plate causing the average aortic pressure to increase), the average aortic pressure would increase slowly and then suddenly jump to a much higher mean pressure. As the atrial pressure approached another critical pressure, the atrium would collapse and the average aortic pressure would jump up. This problem was solved by physically restricting the atrium size with a rod not allowing the atrium to form a fully inflated sphere, thus forcing it to operate with lower volume where it could not exhibit bi-stable modes.

The third nonlinearity manifested itself in a similar way as the atrial problem. The mean pressure would suddenly jump from one pressure to another. The difference, in this case, was the magnitude of the average pressure jump, approximately 5-7 mm Hg. The cause of this nonlinearity was traced to the construction of the resistor plate. The manufacturing process introduced peaks and valleys in the plate, causing it to deviate from being a flat surface. Once identified this problem was easily corrected by altering the process by which the resistor plate was manufactured.

In the final orbital version of the mock circulation system, these nonlinearities have been recognized and eliminated. It is anticipated that the elimination of these nonlinearities will enable the window around 95 mm Hg, to be narrowed to less

than  $\pm 0.5$  mm Hg. The proposed experiments will contribute to the understanding of the mechanisms of cardiovascular adaptation to weightlessness and help to establish the methodology for the investigation of the long-term effects of weightlessness during space flight on board the Space Station and during interplanetary flight. These mechanistic findings of cardiovascular adaptation may also contribute to the development of effective cardiovascular countermeasures for crew members upon returning to Earth and the fundamental effects of gravity on cardiovascular function.

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