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Performance Analysis of Pneumatically Driven Blood Pumps[1]

RAYMOND J. KIRALY, GORDON B. JACOBS, JORGE URZUA, AND YUKIHIKO NOSÉ

Department of Artificial Organs, Division of Research, Cleveland Clinic Foundation, 9500 Euclid Avenue, Cleveland, Ohio 44106

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A simplified analysis of the performance characteristics of pneumatically actuated artificial hearts was made. The analysis was based on the assumption that inertia forces predominated during the filling and emptying of the ventricles. The results qualitatively predicted *in vitro* characteristics and allowed control criteria to be based on physical principles.

INTRODUCTION

Currently a good deal of effort is under way in the research of artificial hearts as total replacement and as assist devices. The principal tool in this research is the pneumatically driven artificial heart implanted in calves. The pneumatically driven devices are used mainly because they are inexpensive and simple to operate and control. In addition the first clinically used left ventricular assist devices will probably be pneumatically driven pumps (NHLI, 1974). One of the first steps in the successful use of the pneumatically driven ventricle is to understand the characteristics of its operation. While virtually all artificial heart developers employ *in vitro* performance testing, few have attempted explanations for the performance characteristics. Phillips *et al.* (1974) have done complete performance mapping and have postulated dimensionless parameters for data display. Walker *et al.* (1974) have attempted dimensional analysis to define the performance parameters. We have taken a unique approach in attempting to analyze the physics of operations in order to arrive at the general performance characteristics. Although we have previously presented aspects of this analysis (Kiraly *et al.*, 1974a, 1974b; Urzua *et al.*, 1974), this article is the first complete description of our understanding of pneumatically driven ventricles and the ramifications affecting control of the devices.

ASSUMPTIONS AND ANALYSIS

Figure 1 shows a typical pneumatically driven ventricle. The air and blood chambers are separated by a flexible diaphragm. When the air pressure (P_a) is reduced below the venous pressure (P_v), the inlet valve opens and blood fills the

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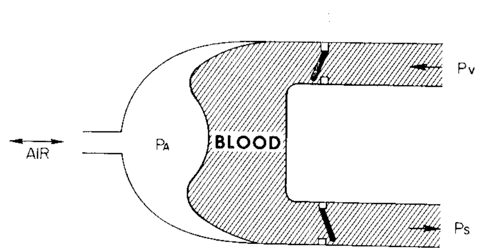

FIG. 1. Schematic representation of an air driven blood pump. P_a, P_v, and P_s are air, venous, and system pressures, respectively.

ventricle. When the air pressure is increased above the systemic pressure (P_s), the blood is ejected from the heart into the artery. Figure 2 shows the idealized pressures assumed for the purpose of the analysis. The air pressure waveform is assumed to be a square wave. The systemic pressure and the venous pressure are assumed to be constant during systole and diastole, respectively. During systole of the artificial heart, the air pressure is increased above P_s, and the net actuating pressure is, therefore, ΔP_S. During diastole the air pressure is reduced below P_v, and the net actuation pressure for filling is ΔP_D.

To obtain an understanding of the characteristics of the pneumatically driven ventricle, a simple dynamic analysis was performed involving three principal assumptions. The first is that the filling and emptying processes are independent. This means that the variables affecting filling do not affect the emptying process, and vice versa. Therefore, during systole, the time of systole (S) and the pressure difference (ΔP_S) are the only variables affecting the performance. Similarly, during diastole only ΔP_D and diastolic time (D) are the significant variables. The second assumption is that pressure differences only are significant. Pressure level itself, while having physiological significance, it not applicable to the performance of the ventricle. The third assumption is that forces are converted into fluid ac-

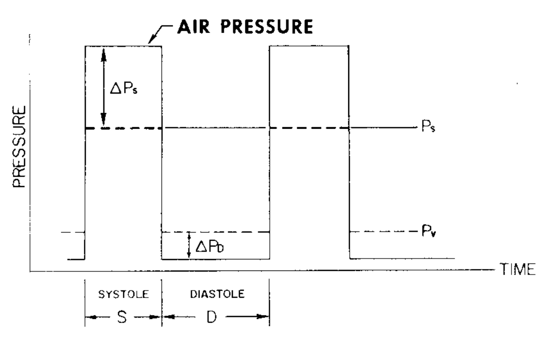

FIG. 2. Pressures assumed during the pumping cycle for purpose of analysis. Air pressure is a square wave; systemic and venous pressures are constant. ΔP_S is the difference between the air and systemic pressures during systole. ΔP_D is the difference between the air and venous pressures during diastole.

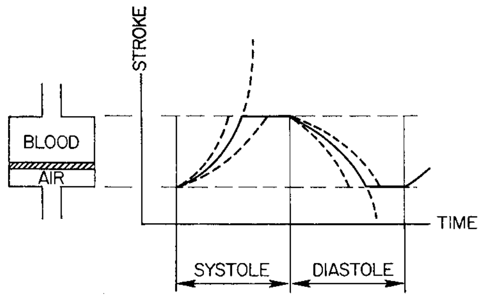

FIG. 3. Ventricle motion follows a square curve during filling and emptying.

celeration. To obtain a product-type solution, both resistance and inertia terms could not be used together. Therefore, the assumption was made that the inertia forces are predominant, and the resistances are neglected. During each portion of the cardiac cycle, the force on the fluid, and consequently the acceleration, is proportional to the ΔP existing during that time. Under constant acceleration, the displacement is proportional to the second power of time. Therefore, the stroke volume is proportional to the product (ΔP) $(\text{time})^2$:

$$\text{Stroke volume} \propto (\Delta P)(\text{time})^2. \tag{1}$$

The cardiac output (Q) is equal to the product of stroke volume and heart rate. Heart rate (HR) is equal to the reciprocal of the cardiac period:

$$HR = 1/(S + D), \tag{2}$$

where S is the period of systole and D is the period of diastole in each cardiac cycle. The diastolic ratio (R_D) is defined as the fraction of the cardiac cycle used

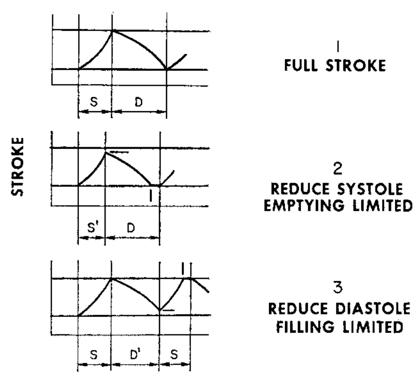

FIG. 4. Effects of varying the systolic and diastolic durations. S and D are durations required to complete a full stroke during systole and diastole, respectively. $S^1 < S$ and $D^1 < D$. Reducing the duration of systole results in incomplete emptying and stroke volume will vary with ΔP_S. Likewise reducing the duration of diastole results in incomplete filling and stroke volume will vary with ΔP_D.

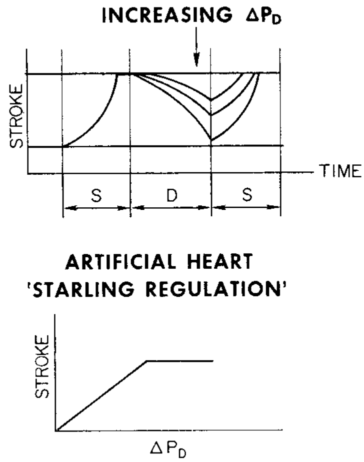

FIG. 5. In filling limited mode, the stroke volume is directly proportional to ΔP_D similarly to the Starling regulation for the natural heart.

for diastole:

$$R_D = D/(S + D); \tag{3}$$

likewise,

$$R_S = S/(S + D). \tag{4}$$

Combining these terms yields the simple function

$$Q = K(\Delta P)(R^2)/HR. \tag{5}$$

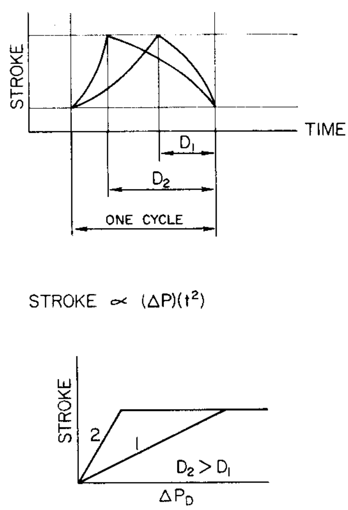

FIG. 6. As the time of diastole increases, less filling pressure difference is required for the same stroke volume. Sensitivity to filling pressure is increased with increased time for diastole.

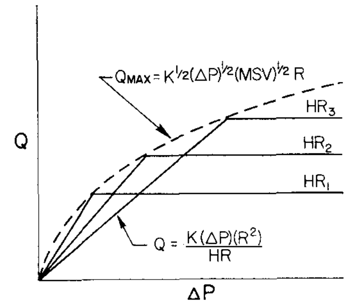

FIG. 7. Family of performance curves showing effects of operating variables. $HR_3 > HR_2 > HR_1$ Higher flows (Q) are attainable with higher heart rates (HR) but at a sacrifice in flow/pressure sensitivity in the operating region.

This equation is applicable to both systole and diastole using the proper consistent subscripts on both (R) and (ΔP). For convenience, the proportionality constant, K, is assumed to be the same for systole and diastole. Since there are two portions to the cardiac cycle, the cardiac output could be dependent on either the filling conditions or the emptying conditions, depending upon which portion of the cardiac cycle has the lesser value for the product $(\Delta P)(R^2)$. If the product $(\Delta P)(R^2)$ for diastole is less than that during systole, we would consider the heart to be filling limited. In this situation, during each systole all of the blood is emptied, while during each diastole the ventricle is only partially filled. Similarly, the device would be emptying limited if the ventricle is completely filled during diastole, but only partially emptied during systole. Each device has a maximum stroke volume (MSV) obtainable. HR can be eliminated from Eq. (5) by substituting

$$HR = Q_{\max}/MSV \tag{6}$$

to yield

$$Q_{\max} = (K)^{\frac{1}{2}}(\Delta P)^{\frac{1}{2}}(MSV)^{\frac{1}{2}}R. \tag{7}$$

This equation defines the locus of the points where maximum cardiac output is reached.

Figure 3 illustrates the mechanics of operation. Schematically, on the left are shown the air and blood in the ventricle with a frictionless piston between the two. The piston-cylinder model is used for convenience to visualize the displacement and motion in the ventricle. The curves trace the motion of the piston as a function of time. During systole the motion of the piston follows a (time)^2 curve shown by the solid line which is terminated when the piston reaches the end of its travel. During diastole the piston will move down on a (time)^2 curve as shown by the solid line until the piston reaches the end of its travel. The dashed lines indicate the motion that would occur if the pressure differences were increased or decreased during the systole and diastole portions of the cardiac cycle. The solid curves in Fig. 3 shows that the device is operated at maximum stroke volume since the piston is moving through its full travel.

Figure 4 illustrates the options that are available in operating the device. In the first case full stroke is made during systole. The ventricle is completely emptied

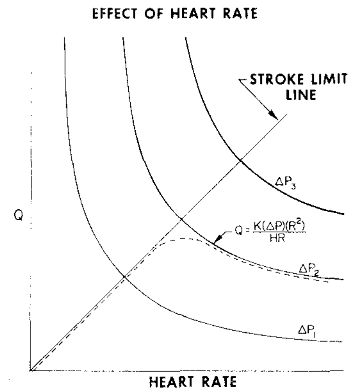

FIG. 8. Same relationships as in Fig. 7. At low heart rates and adequate ΔP full stroke is made and the flow is proportional to heart rate. At high heart rates, ΔP is insufficient for complete filling (or emptying) and the flow will decrease with increased heart rate.

and, simultaneously with the point at which the piston reaches full travel, diastole begins, and the heart is filled by the time systole again occurs. In the second case systole is reduced so that S^1 is less than S in the first case, and consequently the ventricle is not completely emptied because not enough time is available. Diastole is initiated with a partially filled ventricle, and more than enough time is available during diastole to completely fill the device. Since all of the blood in the ventricle is not completely emptied, this is defined as an "emptying limited" situation. Note that in this case altering ΔP during systole will affect the stroke volume and cardiac output. Referring now to the third case, there diastole is reduced so that D^1 is less than D in the above examples, the first systole completely empties the ventricle. The ventricle is only partially filled during diastole and is completely emptied prior to the end of each subsequent systole. Case 3 is the "filling limited" condition defined by the situation where all of the blood is emptied during each systole but the ventricle is only partially filled during each diastole. The stroke volume is therefore equal to the end diastolic volume. In this case, note that the changes in the ΔP during diastole will affect the cardiac output. Since this phenomenon is inherent in the natural heart, it is desirable to have this condition in the artificial heart as well.

Figure 5 illustrates the effect of ΔP on cardiac output. The upper diagram shows that during each systole all of the blood that is in the ventricle is completely emptied, while during diastole, the ventricle is only partially filled. Note that as ΔP_D increases, the degree of filling increases. Plotted in the lower diagram of Fig. 5 is the actual stroke as a function of ΔP_D, which is linear; consequently, at a given heart rate, the cardiac output is directly proportional to ΔP_D. This is the "Starling Regulation (Guyton, 1971) for the artificial heart. Note that the point is reached where full stroke is made, and any increase in ΔP_D above this value does not increase the stroke and cardiac output.

Figure 6 indicates how the sensitivity to ΔP_D can be altered. Shown in the upper diagram of Fig. 6 are two different periods of diastole during the same cardiac period. In the first case, where diastole is relatively short, D_1, ΔP_D must be high in order to completely fill the heart during D_1. If the period of diastole is increased to D_2 by employing a short systole, a relatively low value of ΔP_D is required to completely fill the ventricle. Since the stroke volume, and hence the cardiac output, is proportional to the product of ΔP and $(\text{time})^2$, doubling the period of diastole from D_1 to D_2 would result in four times the sensitivity to ΔP_D as indicated in the lower diagram. Therefore, by adjusting the period of diastole, the sensitivity to ΔP_D can be changed. Since ΔP_D is the difference between the atrial pressure and the air pressure in the ventricle during diastole, the cardiac output can be plotted against the atrial pressure for a constant value of the air pressure during diastole. This is essentially how pneumatically driven ventricles are operated in order to obtain Starling regulation. It is important to stress once again the fact that the sensitivity to atrial pressure only occurs in the filling limited mode of operation. If the device is in the emptying limited condition, it will exhibit a similar sensitivity to arterial pressure with essentially no sensitivity to the atrial pressure. It is important to keep this principle in mind.

Referring now to Fig. 7, we can examine a family of curves showing the relationship between the cardiac output, ΔP, and heart rate, while maintaining constant systolic ratio. Note that at low rates, HR_1, a high sensitivity to ΔP is exhibited; however, the highest possible cardiac output is limited by the product of the maximum stroke volume and heart rate. If the heart rate is increased, a somewhat lower sensitivity is obtained, while a higher maximum output is possible. The dashed curve shows the locus of the points where maximum cardiac output is reached. The information to be gleaned from this figure is that ventricles having as large a maximum stroke volume as anatomically feasible and operated at a heart rate as low as consistent with the maximum cardiac output required by the animal, will result in the highest sensitivity to filling pressure. This is desired for the artificial ventricle, in order to mimic the inflow sensitivity of the natural heart (Nosé *et al.*, 1967).

Figure 8 shows a plot of the same variables, this time plotting the cardiac output as a function of heart rate. Since the cardiac output is inversely proportional to heart rate for constant ΔP and constant R, cardiac output can be increased by increasing ΔP (ΔP_1 being the lowest and ΔP_3 the highest shown on this figure). Cardiac outputs above the stroke limit line are not possible, since this line is defined by the product of the maximum stroke volume and the heart rate. A typical performance curve might follow the dashed line shown in this figure. The information to be obtained from this figure is that if cardiac output is found to increase with heart rate, then full stroke of the heart is being made and the heart will not exhibit any sensitivity to pressure difference either in filling or in emptying. For the device to operate with Starling regulation (sensitivity to ΔP), it must exhibit decreasing output with increasing heart rate.

Figure 9 illustrates the effect of the systolic and diastolic ratios on the cardiac output and also shows a number of interesting features. Filling limited conditions exists to the right of the peak shown in this curve, generally at low diastolic ratios, while emptying limited conditions exist to the left of the peak at low systolic ratios. The curves on this figure are drawn for constant ΔP's. The to left of

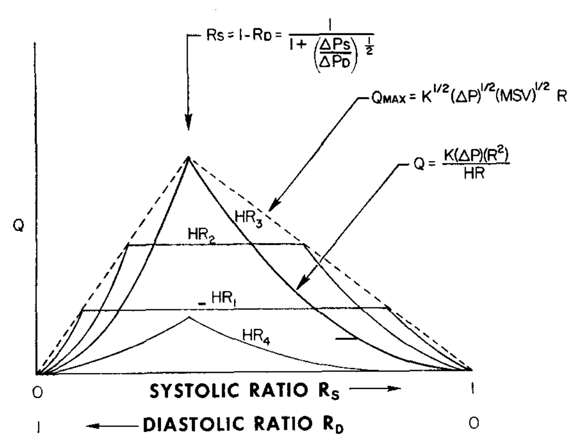

FIG. 9. Effect of systolic ratio. Subscript S is systole, D is diastole, $HR_1 < HR_2 < HR_3 < HR_4$. Starling regulation occurs only to the right of the peak and at submaximum capacity. Emptying limited conditions exist on the left of the peak. The systolic ratio at the peak is determined by the ratio of the ΔP's only in the equation shown.

the peak, Q is a function of ΔP_S and R_S, whereas to the right of the peak the curves refer to ΔP_D and R_D. For example, on a typical curve shown at HR_1 and low R_S, Q is shown to be a function of the $R_S{}^2$ until maximum cardiac output is reached as defined by the dashed line. Increasing systolic ratio at HR_1 does not affect cardiac output until systolic ratio is increased sufficiently so that the device becomes filling limited. Q then decreases with further increases in R_S. If the heart rate is increased to HR_2, higher maximum output is obtainable proportional to the heart rate. HR_3 indicates the unique condition where at a particular R_S the ventricle makes a complete stroke with no dead time in the cardiac cycle; that is, the heart just completes emptying when diastole starts and starts systole again just at the point where the heart is completely filled. An increase in heart rate to HR_4 will reduce the cardiac output in any portion of the performance map. The only way to obtain higher output in this case would be to increase the ΔP. Note that by equating the Q during diastole to the Q during systole results in defining the peak point as a function of the ΔP's only. The normal operating region for the artificial ventricle in order to obtain Starling regulation is on the right-hand side of the peak for filling limited conditions.

DISCUSSION

The analysis and performance curves that have been presented were based on the assumption that inertia forces predominate in the filling and emptying of the artificial ventricle. Since the total impedance to flow is made up of inertia, capacitance, and resistance, we would expect actual performance curves to deviate somewhat from those presented here. The purpose of this analysis was not to predict performance quantitatively, but to show the performance characteristics and to explain the operation of pneumatically driven ventricles under varying conditions. The analysis also indicated some principles of operation which were im-

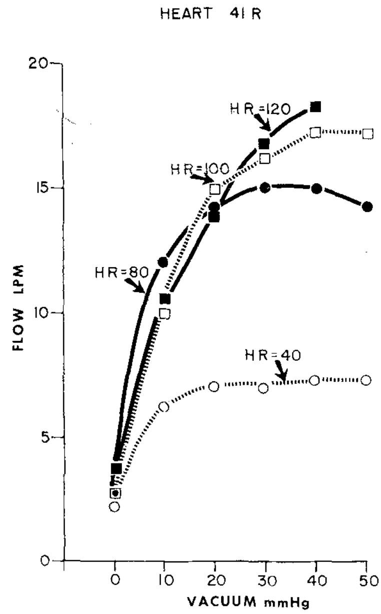

FIG. 10. *In vitro* performance data for artificial ventricle showing sensitivity to filling pressure difference. For convenience in testing, atrial pressure is maintained constant and diastolic air pressure varied. P_S = 40 mm Hg, P_a = 200 mm Hg, P_v = 4 mm Hg, R_S = 0.40.

portant in obtaining the sought-after Starling-type regulation in the artificial heart. These can be summarized by stating that the best performance can be obtained with the largest maximum stroke volume anatomically feasible for fit in the animal and operation at the lowest heart rate consistent with the maximum cardiac output required. Also, a low systolic ratio with adequate systolic air pressure, so that the ventricle is emptied on each stroke of the heart, is necessary. This will then ensure that the device will be filling limited and be sensitive to the inlet or atrial pressure while holding the diastolic air pressure constant. Also, by variation in the diastolic air pressure, the Starling curve can be shifted to the right or the left in order to control the atrial pressure operating line. Atrial pressure changes in themselves will then allow the output to move up and down this curve about the set point. This means that a family of Starling curves can be drawn, each displaced from each other by variations of the diastolic air pressure. Any one of these curves can be selected by the proper selection of diastolic air pressures. The slope of the curve, however, will remain constant, dependent on the diastolic ratio and the heart rate.

The results of this analysis have proved very useful to us, first of all in reducing the amount of *in vitro* testing in order to obtain a complete performance map. Second, only a few data points are required once the shape of the curve is known. Knowing the interrelationship of the operating variables, one can then limit testing to the region of interest such as the filling limited conditions. The analysis also removes some confusion in examining data without an understanding of what is actually happening. For example, in one portion of Fig. 9 one observes that initially increasing output occurs with increasing heart rate, while increases in heart rate eventually reduce the output. This analysis has accomplished the purpose of lending some insight into the physics of operation of the pneumatically driven ventricles.

Figure 10 shows an *in vitro* performance curve for a ventricle used as a right heart. For simplicity of testing, the atrial pressure was maintained constant while the diastolic air pressure was varied. The similarity to the analytically derived curves in Fig. 7 is apparent. Although insufficient data at the heart rate of 40 prevent determining the initial slope, the decreasing sensitivity with increasing heart rates is observed for the higher heart rates. Figure 11 shows the same data plotted as flow versus heart rate to show the similarity to the predicted curves of Fig. 8. The close agreement of the predicted curves with test data of our own, as well as with data from other investigators, indicates that inertia may be more significant than generally regarded. It is also remarkable that the predicted systolic ratio, R_S, for maximum flow given by the equation in Fig. 9 has generally agreed with much of the published data we have seen, obtained with a variety of devices and mock circulatory systems. For example, in the data of Phillips *et al.*, R_S for peak flow was measured at 0.27 while the analysis predicts a value of 0.26 for their conditions of air and blood pressures.

The principal deviations in actual data from the analytically predicted curves depend somewhat on the characteristics of the *in vitro* test setup that is used to measure the performance. In some mock circuitry systems, the use of long lines of relatively small diameter would tend to increase the effective inertia. Also, the air pressure is generally not a square wave, and ΔP is not constant over the entire

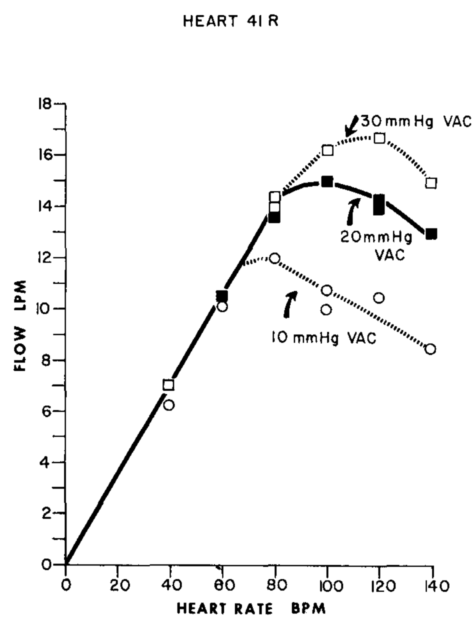

FIG. 11. *In vitro* performance, same data as in Fig. 10, plotted as function of heart rate.

portion of the cardiac cycle. Variations in the atrial pressure during filling of the device also changes ΔP_D. This is especially true in the *in vivo* experiments where atrial pressure variations are significantly higher than most *in vitro* setups and higher than measured with a normal heart. Leakage of the heart valves and regurgitation reduce cardiac output. In this case, the assumption that the filling and emptying processes are independent no longer holds, because valve leakage is a function of the pressure differences across the valve. Another factor contributing to the effect of pulse rate on output is the fact that a dead time exists in the cardiac cycle between the time when the air pressure is changing from diastole to systole and vice versa. Whenever the air pressure is between the venous pressure and the systemic pressure, no output or input of blood occurs. Therefore, a pneumatic driving system having time lags and significant resistances in the air lines would show a decrease in output with increasing heart rate because, as the heart rate increases, the percentage of dead time in the cardiac cycle would also increase. Therefore, the actual situation includes the inertia, capacitance, resistance, and the dead time effects in varying combinations depending on the exact setup. However, the simplified analysis presented here was felt, from our own experience, to indicate the performance characteristics in enough detail so that an understanding of the principles and proper operating techniques could be determined.

In summary, a very simple analysis was performed to obtain performance characteristics of pneumatically driven artificial hearts. To obtain a simple solution several assumptions were required without which the analysis would have lost its usefulness. In spite of what seem to be oversimplifications, qualitative and quantitative results adequately predicted test results. The analysis also allowed performance characteristics to be understood sufficiently so that control of artificial hearts could be based on the physics of operation.

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