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MICROPROCESSOR CONTROL OF A VENTRICULAR VOLUME SERVO-PUMP

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Measurement of the instantaneous pressure–volume relationship of the left ventricle is fundamental to the study of ventricular mechanics. In order to effectively investigate this relationship, it is necessary to vary and control the time course of ventricular volume change in a variety of prescribed manners. In the past, we used an analog circuit to generate command signals for a servo-pump system which controlled ventricular volume. The use of analog control limited the variety of volume waveforms which could be generated. To overcome this limitation, we developed a new system in which the servo-pump is controlled by an inexpensive microprocessor based computer, capable of generating an unlimited repertoire of volume waveforms. The computer system also made possible the use of adaptive control to increase the system fidelity. Finally, such a system provides for ease of adjustment to new hardware, should future research require it.

Keywords — *Servo-pump, Microprocessor, Ventricular pressure–volume relationship.*

INTRODUCTION

A number of indices of "cardiac contractility" have been proposed by physiologists to characterize the functional state of the ventricle (1). However, because it is difficult to control and measure ventricular volume in patients and experimental animals, most frequently used indices, such as dp/dt, $dp/dt/p$, and V_{max}, are derived from left ventricular pressure curves alone. In general, these indices are sensitive to the inotropic state of ventricle, but are also sensitive to preload and afterload. This makes the application of these indices somewhat limited.

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In the early 1970's, Suga *et al.* (10) and Suga and Sagawa (7) rekindled the interests of cardiologists in the pressure–volume relationship of canine left ventricle by showing the usefulness of this relationship in characterizing the ventricular inotropic state. Suga and Sagawa (7) controlled the volume of isolated canine ventricles by putting a balloon in the left ventricle which was connected to a passive volumetric chamber. For further investigation of the pressure–volume relationship, they (8) developed an active servo-pump to control ventricular volume and they (6,9) obtained more detailed information about the instantaneous relationships between the systolic pressure and systolic volume. In these studies, the command signal to the servo-pump was generated by an analog circuit by charging and discharging a capacitor with different time constants. Although the patterns of the command signal were limited, the pump clearly demonstrated the usefulness of controlling the instantaneous ventricular volume for the study of ventricular mechanics. Other investigators (2,3,4,5,11) also recognized the importance of instantaneous ventricular volume control as a means to study the ventricular mechanics. Covell *et al.* (2) developed a servo-pump to impose an isotonic contraction on the canine left ventricle. Templeton and Nardizzi (11) imposed sinusoidal volume perturbation on isolated canine ventricles and measured the ventricular pressure response to determine the elastic and viscous properties of the ventricle. Schiereck and Boom (5) developed a servo-pump which could impose step changes in ventricular volume on isolated rabbit hearts. Using this device they measured active stiffness of left ventricle. Hunter *et al.* (3) analyzed response of isolated canine left ventricles to a flow pulse which is fairly brief in duration (from 35 to 50 msec) using a servo-pump designed by Janicki and his associates (4). They pointed out the importance of viscous properties of the ventricle near the end of systole in addition to the importance of its elastic properties. All these investigations used a variety of time dependent changes in ventricular volume to study the unique features of cardiac muscle. The volume servo-pump system previously developed in our laboratory (8) could not impose these unusual instantaneous volume changes on the ventricle. To overcome this limitation and extend the scope of our analysis through the use of a greater variety of volume command signals, we have developed a new servo-pump system which is controlled by a microprocessor.

The design goals set for the new servo-pump system were (i) versatile volume waveform generation, (ii) servo-pump hardware improvement, (iii) coronary perfusion pressure control, and (iv) pacing signal generation. The most important of these is the capability to generate variable volume waveforms. Theoretically a microprocessor, which is to be used as the controller, can generate a volume command signal with any time pattern. However, the nonlinearity and dynamic characteristics of the servo-pump hardware (piston pump and linear motor) limit the performance of the system as a whole. Specifically, if the pump hardware poorly follows the command signal, the unlimited ability of the microprocessor to generate complex command signals is useless. We have utilized the microprocessor to compensate for the limited performance of the system's hardware and have thus improved the overall system fidelity.

In addition to the generation of the volume command signal, it is necessary that the controller produces a ventricular pacing spike which was appropriately timed to the volume command signal. Finally, the controller must generate a command signal for a servo perfusion system which regulates the perfusion pressure of the coronary arteries.

All of these multiple real-time tasks can be done relatively easily by an inexpensive microprocessor based computer.

SYSTEM CONFIGURATION

The overall system is comprised of two major components: the controller and the servo-pump hardware. The servo-pump hardware configuration is similar to the one described by Suga and Sagawa (8). A schematic reproduction of this hardware is given in Fig. 1. For detailed information on the system's components, the reader is referred to the appendix of Ref. 8. A latex balloon, over which the left ventricle (LV) is placed, connects to a rolling diaphragm cylinder (BFP) via a connecting tube (*C*). AV is an air vent through which the system is primed with water (*W*), without leaving air bubbles. The piston in the cylinder is connected to the plunger (*P*) of a linear motor. Both the cylinder and the linear motor are mounted on a sturdy frame (*F*). The position of the piston is sensed by a linear variable displacement transducer (LVDT). This signal is subtracted from the reference signal (REF INPUT) (the desired left ventricular volume) to produce an error signal for the volume servo-control. This error signal is amplified by the power amplifier (PW) which drives the linear motor.

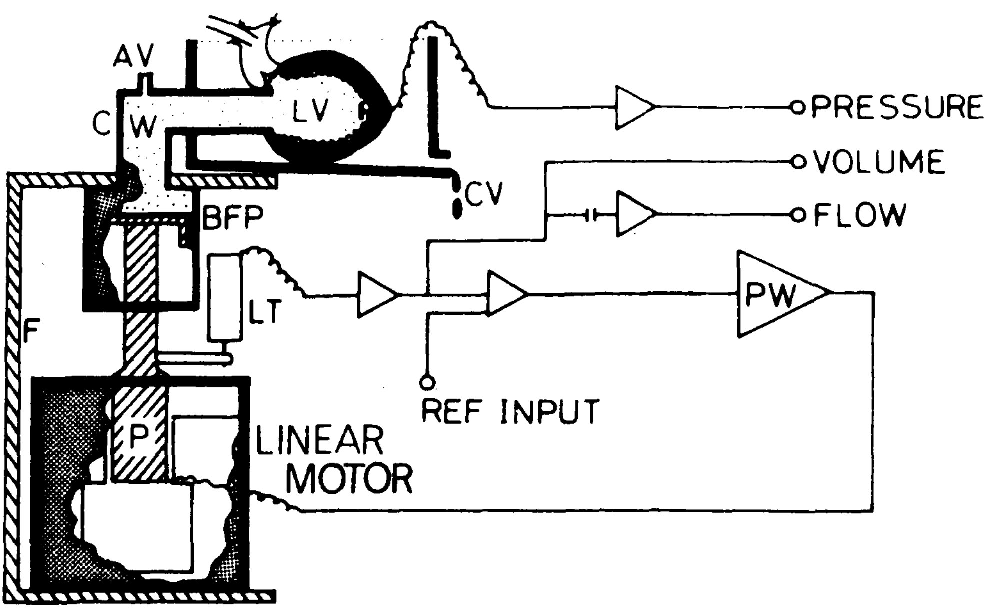

FIGURE 1. Schematic diagram of volume servo-pump hardware. REF, volume command signal; CV, coronary venous blood. See text for detailed explanation. [From Suga and Sagawa (8) with permission.]

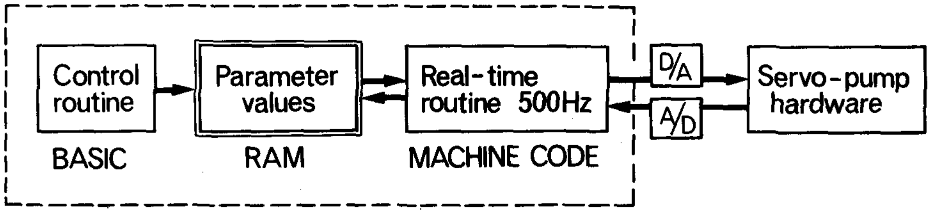

FIGURE 2. The system consists of two sections: the microprocessor based controller and the servo-pump hardware. In the controller, there are two sections of control software. The real-time routine feeds the volume command signal into the servo-pump hardware with a specially designed *D/A* converter and at the same time monitors the volume change through an *A/D* converter every 2 msec. The control routine (left block) allows the experimenter to communicate with the real-time routine in order to change stroke volume, end-systolic volume, end-diastolic volume, waveform, pacing rate, and coronary perfusion pressure.

The controller is built around the Radio Shack microcomputer which uses a Z-80 microprocessor (ZILOG). Specially designed analog to digital (*A/D*) and digital to analog (*D/A*) converters have been interfaced with the computer. The microcomputer executes a real time routine and a control routine simultaneously. The interactions between the two routines and their relationship to the servo-pump hardware are schematically illustrated in Fig. 2. The overall system configuration is shown in Fig. 3.

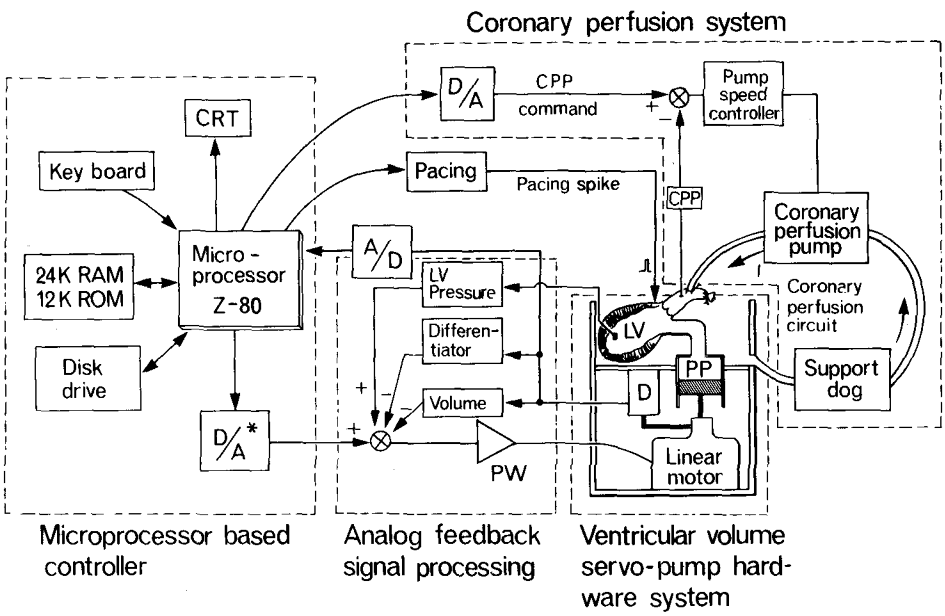

FIGURE 3. *Schematic diagram of the overall system.* Outputs from the microprocessor are the volume command signal (*D/A**), coronary perfusion pressure control command (*D/A*) and the pacing command. Input to the microprocessor is instantaneous ventricular volume. The experimenter communicates with the microprocessor system through a keyboard and CRT. LV, left ventricle; PP, piston pump; D, displacement transducer; PW, power amplifier, and CPP, coronary perfusion pressure.

The control routine is written in BASIC. It generates new waveforms and controls parameter values which are used by the real-time routine. These parameters include heart rate, stroke volume, end-diastolic volume, and end-systolic volume. Information provided by the control routine is stored into specific memory locations. The real-time routine communicates with the control routine through this shared memory space.

The real-time routine is written in machine code to minimize its execution time. It performs the following tasks: (i) provides the volume command waveform to the servo-pump hardware every 2 msec (i.e., 500 Hz); (ii) produces a pacing spike once every cardiac cycle; (iii) provides coronary perfusion pressure command signal once every cardiac cycle; and (iv) digitizes the actual ventricular volume signal which is produced by the piston position sensor.

EJECTION PATTERN GENERATION

One kilobyte of memory is set aside for storage of the volume command signal in the computer. This memory is divided into two sections. The ejecting pattern being used to control the servo-system is stored in one section and is called the current ejecting pattern. The other section is used as a buffer space in which a new ejecting pattern can be stored. While the real-time routine is reading a current waveform, the control routine can generate a new ejecting pattern and store it in the buffer. When the calculation of the new ejecting pattern is complete, the control routine can swap the memory locations and cause the new ejecting pattern to be outputed to the servo-system.

Because the volume command signal is generated by software, there are essentially no restrictions on its complexity. In our experiments, however, we frequently use a family of ejecting patterns which are related in that the durations of ejecting, filling, and isovolumic phases are identical between patterns, but different in stroke volume, end-systolic volume, and end-diastolic volume. In order to minimize the time the computer takes to alter stroke volume, end-systolic volume or end-diastolic volume, we devised a special combination of *D/A* converters which will perform the task without recalculating an entirely new command signal as described in the following paragraph.

If three *D/A* converters are combined as shown in Fig. 4A, the analog output of the circuit will be

$$\text{output} = K_e \cdot [D_e] \cdot [D_s] + K_d \cdot [D_d] \,, \tag{1}$$

where K_e and K_d are reference voltages,

$[D_e]$ is a digital representation (8 bit) of the ejection pattern which changes every 2 msec,
$[D_s]$ is a digital code (8 bit) proportional to stroke volume, and
$[D_d]$ is a digital code (8 bit) proportional to end-diastolic and/or end-systolic volume.

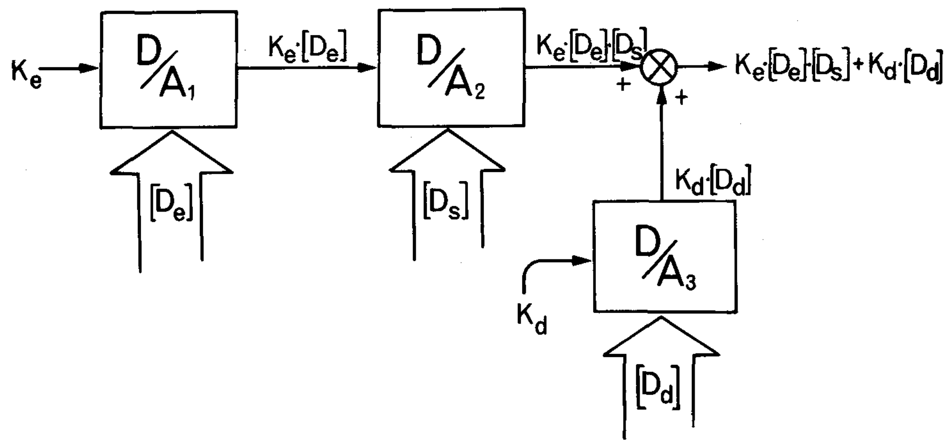

FIGURE 4A. D/A_1 converts the series of digital codes which represent a normalized ejection pattern of volume into the analog signal $K_e[D_e]$. D/A_2 exclusively determines the stroke volume by multiplying the digital code $[D_s]$ by the analog reference signal $K_e[D_e]$. Therefore the analog output of D/A_2 is $K_e[D_e][D_s]$. D/A_3 generates a d.c. voltage corresponding to a digital code $[D_d]$ which determines end-diastolic volume. The analog output of D/A_3 is $K_d[D_d]$. The analog output signals of D/A_2 and D/A_3 are added, resulting in an analog output of $K_e[D_e][D_s] + K_d[D_d]$.

With this special device, the microprocessor can change the stroke volume, end-diastolic volume and end-systolic volume individually or any combinations of them within 10 μsec, after receiving a few instructions. If the microprocessor needs to recalculate the revised volume command signal entirely, it will take at least several seconds, the exact time depending upon the complexity of the ejection pattern. Therefore this combination of *D/A* converters speeds up the microprocessor control system substantially. Furthermore, since the ejecting pattern in the memory is always scaled in a full 8 bit dynamic range, the command signal for a small stroke volume is not limited in resolution. The actual resolution of the volume command signal is 1/256 of the desired stroke volume, which is generally 0.2 ml or less in our experiments.

An example of a computed volume command signal stored in digital form is shown in Fig. 4B. The volume command signal consists of 4 phases. T_0 to T_1 is isovolumic contraction phase; T_1 to T_2, ejection phase; T_2 to T_3, isovolumic relaxation phase; and T_3 to T_4, filling phase. We used parabolas to calculate the volume command signal for both ventricular ejection and filling. The algorithm for calculating this particular waveform was, therefore,

$$V(I) = \begin{cases} 255 & T_0 \leq I \leq T_1 \\ 255\left[1 - \left(\dfrac{I-T_1}{T_2-T_1}\right)^2\right] & T_1 \leq I \leq T_2 \\ 0 & T_2 \leq I \leq T_3 \\ 255\left(\dfrac{I-T_3}{T_4-T_3}\right)^2 & T_3 \leq I \leq T_4 \end{cases} \tag{2}$$

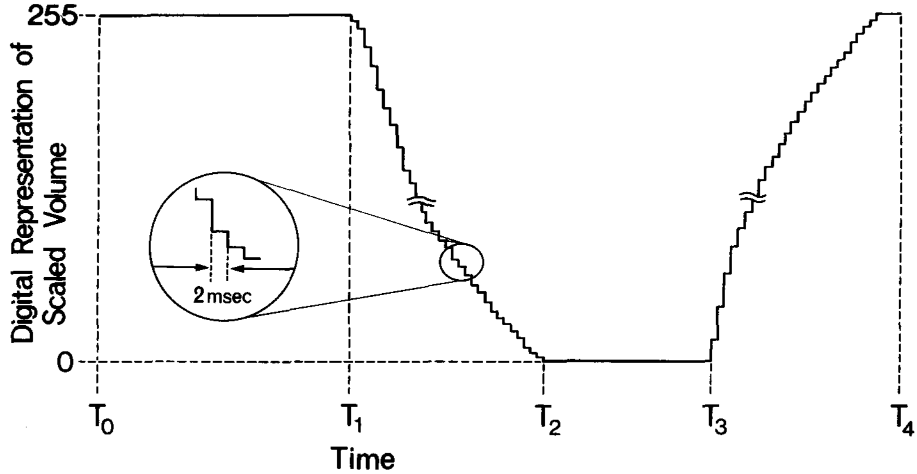

FIGURE 4B. The normalized ejection pattern signal is always fully scaled into 8 bit resolution range independent of the absolute values of stroke volume and end-diastolic volume or end-systolic volume. The time is sliced every 2 msec. T_0-T_1 represents isovolumic contraction phase, T_1-T_2 ejecting phase, T_2-T_3 isovolumic relaxation phase and T_3-T_4 filling phase. The ejection pattern signal is loaded into D/A_1 every 2 msec by the real-time routine program.

where $V(I)$ is the desired volume at time $I*2$ msec after the start of isovolumic contraction and I is an integer. The value of T_1, T_2, T_3, and T_4 can be altered by the experimenter. When these values are changed, the BASIC control routine recalculates and stores a new waveform. An example of actual waveform obtained with the system is shown in Fig. 7 and is discussed in the results section.

PUMP HARDWARE MODIFICATION

We have also modified several parts of the pump hardware to improve the performance of the controlled system. Major modifications are as follows.

Linear Motor

The linear motor (Ling Electronics, Model 411) has a pair of stiff elastic plates, so called "spider," which hold and align a coil assembly around a powerful permanent magnet. Because of these stiff spiders, the relationship of the stroke to the electrical current input to the linear motor is highly dependent on the position (thrust) of the coil. To moderate the nonlinearity, we made the spiders more compliant by thinning each of them to one third of the original thickness. This modification expanded the linear operating displacement range of the piston pump to a considerable degree.

Displacement Transducer

We used a linear variable differential transformer (LVDT) (Transtek, 244-000) as a displacement transducer to feedback the piston position to the volume control system. The transducer was coupled to the plunger of the linear motor via a side arm (Fig. 2). Because the side arm of the original version was slender and compliant, it limited the frequency response and often caused a small dead zone in displacement measurement. We have replaced the side arm with a very rigid one to eliminate these problems.

Analog Feedback

The major negative feedback signal is the ventricular volume signal measured by the displacement transducer (F_1 in Fig. 5). To improve the dynamic response and stability of the hardware, the velocity of the volume change is used as an additional derivative negative feedback signal (F_2 in Fig. 5).

As the ventricle contracts, the increased pressure in the balloon tends to push the linear motor below the desired position. The time-varying interventricular pressure therefore acts as a disturbance to the servo system. Because the open loop gain of the servo-pump system is finite (gain = 10), this disturbance can cause a significant decrease in fidelity of the servo-pump. We therefore added a third signal proportional to the intraventricular pressure as a positive feedback which tends to increase the current output of the power amplifier, thus helping the pump withstand the pressure imposed on it (F_3 in Fig. 5). We owe this pressure feedback idea to Dr. Hiroyuki Suga.

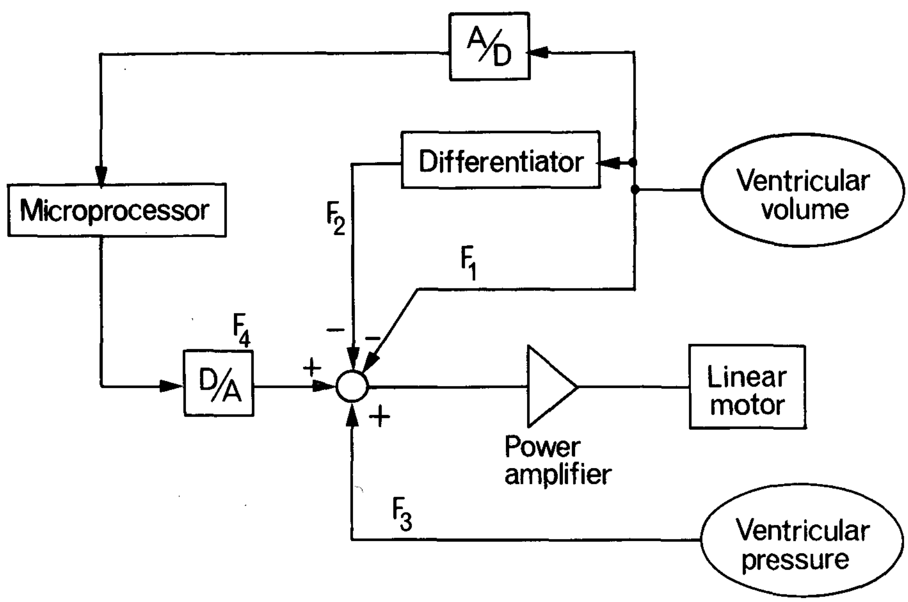

FIGURE 5. Block diagram of multiple feedback controls. F_1, ventricular volume represented by piston position signal is used for a proportional negative feedback control. F_2, the rate of change of volume signal is used for a derivative negative feedback control. F_3, the intraventricular pressure signal is used for a local positive feedback control. For an adaptive control, instantaneous volume signal is digitized (F_4) and the microprocessor compares this signal with the command signal and uses the error signal to modify the *command* signal for the next cardiac cycle.

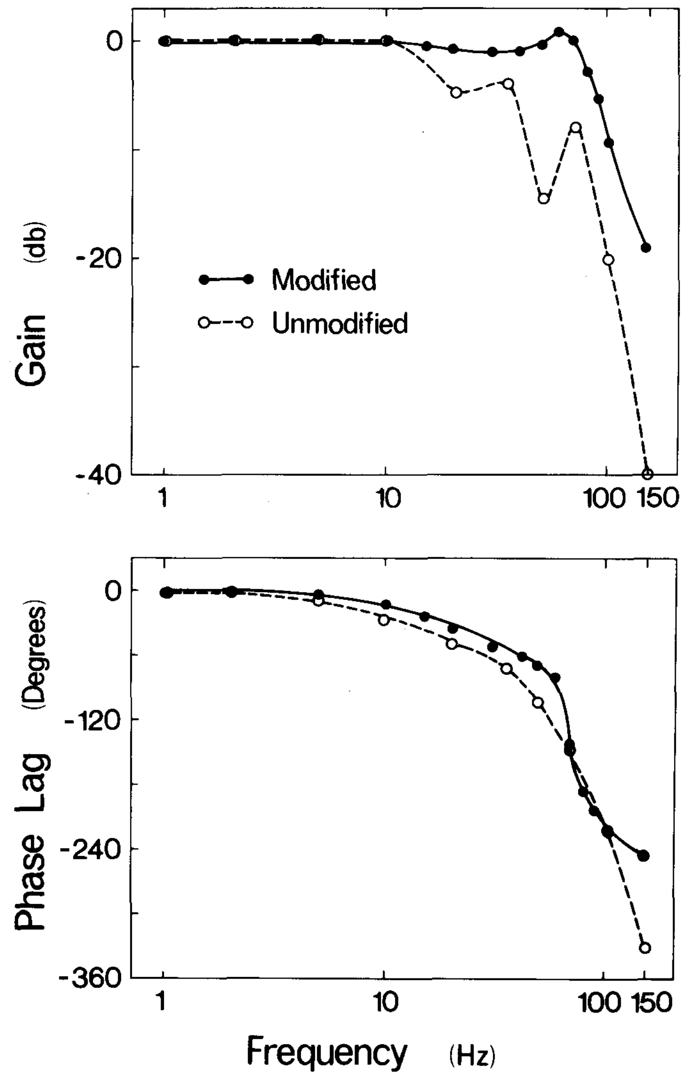

FIGURE 6. Closed-loop Bode plots of the original servo-pump hardware (open circles) and that of the modified servo-pump hardware (closed circles) are shown. Gain was calculated by dividing the output voltage of position sensor (LVDT in Fig. 1) by input command signal voltage (REF INPUT in Fig. 1). Phase lag was calculated by subtracting the input signal phase from output signal phase. Amplitude of input was set equivalent to a stroke volume of 5 ml. Corner frequency of the original servo-pump hardware was about 15 Hz. Corner frequency of the modified servo-pump hardware was about 80 Hz. The Bode plot of the original servo-pump is reproduced from Suga and Sagawa (8).

As an overall effect of these modifications, the servo-pump system became more stable and more linear. The open circles in Fig. 6 show the Bode plot of the closed-loop performance of the servo-pump hardware as reported by Suga and Sagawa (8). The closed circles show that of the servo-pump system after making the modifications. The gain versus frequency curve of the modified servo-pump decreased by 3 dB at 80 Hz as opposed to 15–18 Hz in the original system. As a result of these improved dynamic characteristics, the servo-pump hardware is able to follow the control signal better than before without instability.

COMPENSATION OF HARDWARE BY MICROPROCESSOR

Although the abovementioned modifications of the servo-pump hardware reduced the position-dependent nonlinearity, i.e., the ventricular volume-dependent nonlinearity, it was not completely eliminated. Specifically, if the command signal for stroke volume remained constant, while the end-diastolic or end-systolic volume command was changed, the resultant stroke volume would also vary, the error being as large as 10%. For further improvement of the servo-pump hardware response, we used the microprocessor to implement two types of compensation techniques. One is an open-loop linearization technique and the other is a closed-loop linearization technique.

We first characterized the residual position-dependent nonlinearity by determining the relationship between the command signal and the resultant volume. The microprocessor then generated an inverse function of the nonlinear relation to weight the command signal so that the system response would be linearized (open-loop linearization). This technique took care of about three-fourths of the residual nonlinearity of the hardware.

To eliminate the remaining volume and pressure dependent nonlinearities, we added an additional adaptive feedback loop via the microprocessor (F_4 in Fig. 5). The microprocessor compares the instantaneous actual ventricular volume with the command signal during the two isovolumic phases of the cardiac cycle: $T_0 - T_1$ and $T_2 - T_3$ as shown in Fig. 4B. The difference during each phase is integrated as in Eq. 3.

$$E_{IC} = \sum_{I=T_0}^{T_1} [V_a(I) - V_c(I)]$$

$$E_{IR} = \sum_{I=T_2}^{T_3} [V_a(I) - V_c(I)] \qquad (3)$$

where E_{IC} and E_{IR} are the time integrated errors between the command signal, $V_c(I)$, and the actual volume, $V_a(I)$, during isovolumic contraction and isovolumic relaxation, respectively. $V_a(I)$ is obtained by the computer every 2 msec through the use of the A/D converter shown in Fig. 5, and $V_c(I)$ is supplied to the servo-pump hardware every 2 msec by the computer through the D/A converter shown in Fig. 5. These errors are compared to an allowable error value at the end of each cardiac cycle. If they are greater than the allowable errors, the microprocessor modifies the values of $[D_s]$ and $[D_d]$ in Eq. 1 to minimize the error as in Eqs. 4.

$$[D_s]_{\text{new}} = \begin{cases} [D_s]_{\text{old}} - \dfrac{E_{IR}}{|E_{IR}|} & |E_{IR}| > E_{IR,A} \\ [D_s]_{\text{old}} & |E_{IR}| < E_{IR,A} \end{cases} \tag{4a}$$

$$[D_d]_{\text{new}} = \begin{cases} [D_d]_{\text{old}} - \dfrac{E_{IC}}{|E_{IC}|} & |E_{IC}| > E_{IC,A} \\ [D_d]_{\text{old}} & |E_{IC}| < E_{IC,A} \end{cases} \tag{4b}$$

where the subscripts "new" and "old" refer to the updated and current values of $[D_s]$ and $[D_d]$, respectively, and $E_{IR,A}$ and $E_{IC,A}$ are the allowable errors during isovolumic relaxation and isovolumic contraction, respectively. Because both $[D_s]$ and $[D_d]$ can be changed within 10 μsec, the modification can be done before the beginning of the next cardiac cycle. The compensation sequence will continue until the error is below the prescribed level. It takes a few cardiac cycles to complete the compensation. With this technique, the servo-pump system behaves more linearly despite the nonlinearities of the pump hardware over the wide variation of desired ventricular pressures and volumes. The deviation of the absolute volume from the command signal is within the resolution of the *A/D* converter, which is 0.2 ml.

RESULTS

An example of the pressure and volume tracings obtained with the new system is shown in Fig. 7A. The ejection phase and the filling phase of the volume command were calculated using a parabolic relation between volume and time. The heart rate was set at 120/min, the ejection time was set at 120 msec, and the filling time was set at 100 msec. The end-diastolic volume was set at 25 ml with a stroke volume of 7 ml. Because of the microprocessor based feedback control, the actual ventricular end-diastolic volume and stroke volume are indistinguishable from the values set at the keyboard by the investigator. The resolution in the volume control is up to 0.2 ml which is more than sufficient for our experiments. Figure 7B shows the corresponding pressure–volume loops and three other loops obtained under different loading conditions. The three other loops are obtained by varying end-diastolic volume in steps of 5 ml, while keeping stroke volume constant. The accuracy of end-diastolic volume and stroke volume are well maintained indicating the power of the microprocessor based feedback control.

Figure 8A depicts the ventricular pressure response to a sinusoidal volume pertubation (50 Hz, 0.5 ml) which was superimposed on an isovolumic contraction at a mean volume of 25 ml. The pressure response becomes larger toward the end of systole which reflects time-varying myocardial mechanical properties (11). The same sinusoidal perturbation was then superimposed on an ejecting

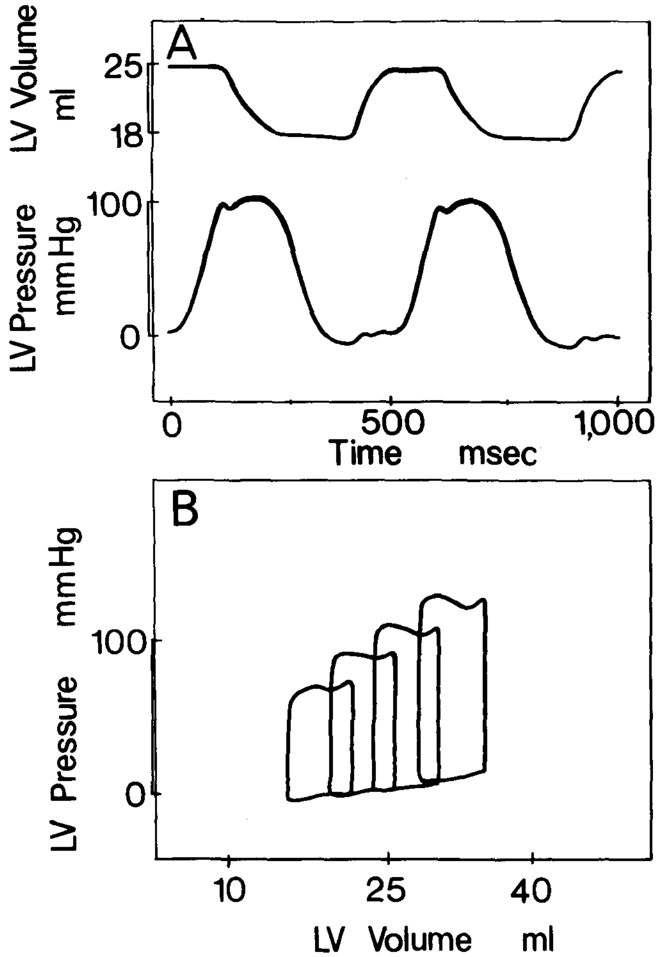

FIGURE 7. (A) An example of a recording of instantaneous left ventricular volume curve (LV volume) and pressure curve (LV pressure). We set end-diastolic volume to 25 ml and a stroke volume to 7 ml by keyboard input. We set heart rate to 120/min, ejection time to 120 msec and filling time to 100 msec. Actual end-diastolic volume generated by the servo-pump is 25 ml with a stroke volume of 7 ml. Actual ejection time is 120 msec and filling time is 100 msec. Therefore, the set parameters for ventricular volume control from the keyboard are exactly realized (within error of reading) by this computerized servo-pump on the isolated canine left ventricle. (B) Examples of ventricular pressure–volume loops. End-diastolic volume was changed in steps of 5 ml while keeping the stroke volume constant at 7 ml. Again, the actual ventricular volume changes exactly follows the set parameter values.

contraction (Fig. 8B). The left ventricular pressure response during diastole is small and then it becomes larger toward end systole.

In Fig. 9, the ventricular pressure responses to flow pulses imposed at four different times of systole for four different beats are shown. The width of the flow pulse is 40 msec and its time integral (volume change) is 2 ml. When a positive flow pulse (ejection) was applied, the ventricular pressure decreased in phase with the flow pulse (Fig. 9A). The amount of the pressure decrease was small when the flow pulse was applied in early systole (the first and second flow

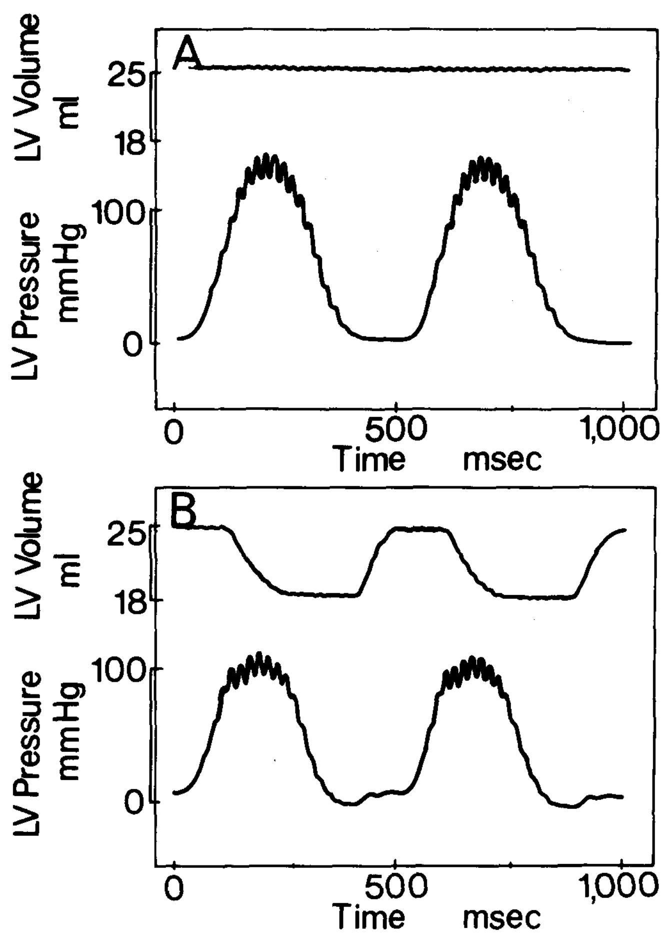

FIGURE 8. (A) Sinusoidal ventricular volume pertubation (50 Hz, $dV = 0.5$ ml) is imposed on an otherwise isovolumically contracting ventricle at the left ventricular volume (LV volume) of 25 ml. Left ventricular pressure (LV pressure) response for this sinusoidal volume pertubation is undetectable during diastole and it becomes larger toward end-systole. The response attenuates again during ventricular relaxation. (B) The same sinusoidal volume pertubation is imposed on a ventricle from the same end-diastolic volume with a stroke volume of 7 ml. The left ventricular pressure response is small during diastole and it becomes larger toward end-systole, a response essentially the same as the isovolumic contraction as shown in panel A.

pulses) and it became larger toward the end of systole indicating the time-varying viscous properties of the myocardium. When a negative flow pulse (injection) of the same magnitude was applied at four different instants in the cardiac cycle of four different beats, the ventricular pressure responses are essentially the same as the positive flow pulse responses except for the polarity (Fig. 9B). These observations are consistent with those reported by Hunter *et al.* (3).

DISCUSSION

The newly developed computerized servo-pump system satisfies all of our current requirements. Because the servo-pump hardware is controlled by the

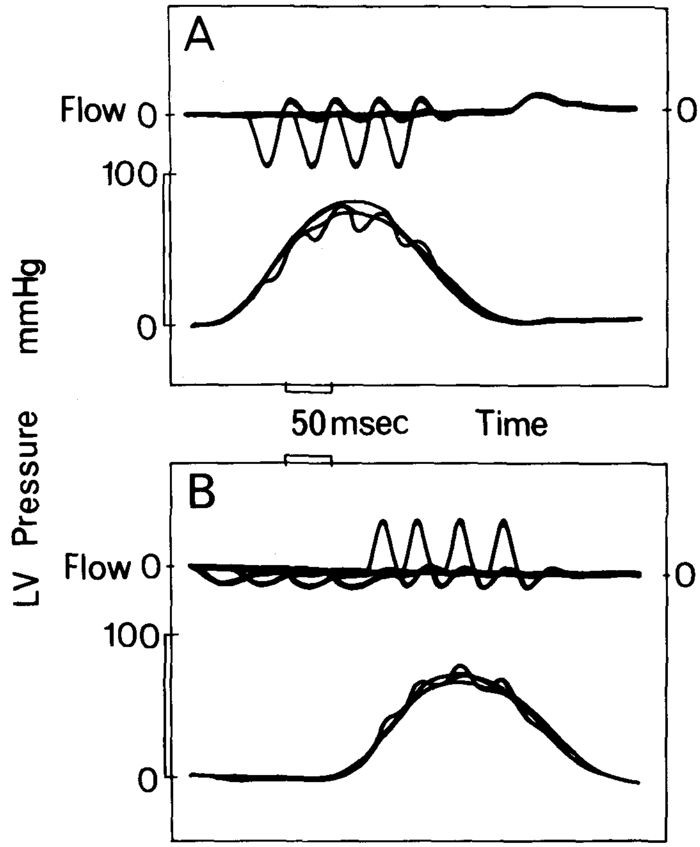

FIGURE 9. (A) Positive volume step of 2 ml over 40 msec was imposed on an otherwise isovolumically contracting ventricle. The figure depicts four superimposed ventricular pressure responses corresponding to those flow pulses imposed at four different instants (first 2 flow pulses: early systole, the third flow pulse: near end-systole and the last flow pulse: after end-systole) in a cardiac cycle. The ventricular pressure response which is in phase with the flow pulse is larger when it is applied near the end of systole (the third flow pulse response). (B) Negative volume step of 2 ml over 40 msec was imposed on an otherwise isovolumically contracting ventricle. The figure depicts four superimposed ventricular pressure responses corresponding to those flow pulses imposed at four different instants (first 2 flow pulses: early systole, the third flow pulse: near end-systole and the last flow pulse: after end-systole) in a cardiac cycle. Although the polarity is opposite that in panel A, the ventricular pressure response which is in phase with the flow pulse is larger when it is applied near the end of systole (the third flow pulse).

computer software in compensatory and adaptive manners, requirements generated by future experimental protocols (e.g., isobaric contraction) may easily be met by this servo-pump system by modifying the control software with a minimum modification of hardware. About 60% of the microprocessor time is spent executing the real-time routine. If it is required that the real-time routine performs an increased number of tasks, a faster microprocessor may be necessary. However for our applications, a single 8-bit microprocessor adequately performs overall control of the servo-pump system.

The estimated cost for the computer hardware including multichannel *A/D* and *D/A* converters is about $1500.

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