A PC based data acquisition system for determination of myocardial preservation during experimental canine studies

by

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INTRODUCTION

A data acquisition system for measuring hemodynamics and myocardial mechanics during a series of experimental surgeries performed on canine subjects was developed. Some experiments required that the data acquisition system be capable of simultaneously displaying and recording the left ventricular pressure waveform (LVP), the left atrial pressure waveform (LAP), single dimensional changes of the left ventricle (segmental shortening (SS)), rate of change of pressure in the left ventricle (dp/dt), mean left atrial pressure (MLAP), and electrocardiograph (EKG). Other studies required the simultaneous display and recording of at least two channels of instantaneous flow, LVP, MLAP, SS, and EKG. Furthermore, the system needed to be sensitive, accurate, versatile, cost effective, easy to calibrate and operate, and safe with respect to operator and subjects.

After careful review and consideration of available systems, the Dataq Instruments AT Codas hardware and software package coupled to a Zenith Z-248 personnel computer with VGA graphics was selected. Signal conditioning units were built and coupled to selected channels at the front-end of the 16 channel system. The system was configured, tested, and used to record data in more than 80 experiments.

LITERATURE REVIEW

Cardiac Surgery

The most common heart operation performed is coronary bypass surgery. This surgical treatment of severe coronary artery disease requires complete or partial cardiac bypass using an extracorporeal oxygenator. Several types of operative procedures are available. Evaluation of prognostic features such as collateral circulation, graft flow rates, and severity and distribution of arteriosclerotic lesions is necessary to compare one type of procedure to another (Parsonnet et al., 1975).

Internal mammary artery graft

The internal mammary artery graft is usually the first choice for coronary artery bypass surgery as it has proven to have the highest long term patency rate. Although a total of two grafts are possible with this procedure, it is usually reserved for cases that require only one bypass. The left internal mammary artery is most often used (Oschner and Mills, 1978). The mammary artery is dissected as a peduncle and freed from the chest wall and diaphragm. The proximal end of the mammary artery is left intact and the distal end is opened and anastomosed to the coronary artery, bypassing lesion sites. This technique offers additional advantages of single anastomosis, artery to artery graft, and no leg wound problems. However, neither right nor left internal mammary artery will reach some distal areas of the coronary circulation such as the terminal circumflex coronary artery (Schimert et al., 1975).

Saphenous vein graft

The autogenous saphenous vein graft is another commonly employed bypass procedure. This procedure is often used in cases that require two or more bypass grafts. The patient's saphenous vein is removed and anastomosed proximally to the aorta and distally to the coronary artery bypassing lesion sites. Although adequate, the long term patency rate is not equal to that of the internal mammary artery graft (Oschner and Mills, 1978).

Coronary-coronary bypass

Coronary-coronary bypass has been introduced for situations in which the internal mammary artery will not reach the distal coronary circulation, adequate lengths of saphenous vein cannot be harvested, or the aorta is not healthy enough for surgery (Rowland and Grooters, 1987). This procedure uses short saphenous vein sections to bypass coronary lesions. Sections are anastomosed to the coronary artery proximally and distally, around blockage sites.

Cardiac ischemia

During open heart surgery, when the heart is bypassed by an extracorporeal oxygenator, a state of temporary global cardiac ischemia exists. During this time the heart may be "stunned" and subsequently temporarily unable to spontaneously contract. The time to recover from temporary ischemia varies from hours to weeks (Buckberg, 1986). Permanent myocardial necrosis may also result from temporary ischemia. The inability to recover immediate spontaneous contractility after global ischemia is not a valid indication of myocardial necrosis, but muscle that can immediately recover contractile function after ischemia must be considered viable (Barnard et al., 1986). Although a great deal is known about the biochemical changes that occur during myocardial ischemia, little is known about the processes which turn reversible "stunned" tissue into irreversible necrotic tissue (Foker et al., 1980).

The levels of adenosine triphosphate (ATP) and creatine phosphate (CP) in the myocardium decrease with ischemia. If the period of ischemia is excessive, ATP levels may never recover and the mechanical function of the heart will be lost (Foker et al., 1980). Nevertheless, since "stunned" myocardium exhibits the same low ATP levels as necrotic tissue, a measurement of low tissue ATP levels does not necessarily signify irreversible myocardial damage (Buckberg, 1986). The measurement of postischemic tissue ATP content as an indicator of myocardial function is controversial (Rosenkranz et al., 1986).

Cardioplegia solutions

During open heart surgery, the electrical and mechanical activity of the heart are stopped by a cardioplegia solution. Different cardioplegia solutions are used; however most contain high levels of potassium. Some have calcium or sodium deprivation or high concentrations of magnesium (Elert et al., 1981). The cardiovascular team headed by Dr. Ronald Grooters, at Iowa Methodist Medical Center, Des Moines, Iowa uses a cardioplegia composed of 30 mEq KCl and 20 mEq $NaHCO^3$ per 1000 ml of 5% glucose solution. The cardioplegia is usually chilled to 4^{o} C and infused into the coronary circulation. The potassium depolarizes the myocardium producing sustained diastole (Wyte, 1983). Cold cardioplegia has reduced myocardial damage during ischemia by slowing the metabolism of the heart and reducing the depletion of ATP in the myocardial tissue (Rosenkrantz et al., 1982).

Many researchers have investigated variations in the cardioplegia as a means to improve myocardial preservation during cardiopulmonary bypass surgery and thus assist functional recovery of the heart from ischemia. Lazar and co-workers (1980) found that rearresting the postischemic heart with a second cardioplegia solution containing L-glutamate resulted in nearly complete reversal of ischemic damage. In studies of working ischemic heart preparations from rats, both a ribose enhanced infusion solution and a low calcium (0.05 mM Ca) with EHNA (an adenosine deaminase inhibitor) and adenosine enhanced infusion solution improved ATP maintenance and functional recovery of the myocardium (Pasque et al., 1982; Humphrey and Seelye, 1982). Warm induction of cardioplegia (normothermic cardioplegia) prior to cold induction cardioplegia, as used with high risk patients, has allowed for safer and more prolonged cardiopulmonary procedures on mongrel dogs (Rosenkranz et al., 1982).

Additional preoperative treatment

Based on a controversial study, Lolley and co-researchers (1979) proposed that the preoperative myocardial glycogen level could be important in myocardial preservation during coronary bypass surgery. Improvement in pre-anoxic cardiac nutrition by administration of a 3-day fat-loading diet and overnight infusion of glucose-insulin-potassium solutions was thought to decrease cardiac arrythmias and hypotension.

To study whether increased myocardial glycogen levels improved cardiac preservation during ischemia, Oldfield and co-researchers (1986) treated patients with a preoperative infusion of 20% glucose in 1 liter of water to which 45 mmol

KCL, 10 units of soluble insulin, and 2000 units of heparin had been added. This solution is referred to as GIK. The solution was infused over 12 hours prior to mitral valve replacement. Although results of the study indicated that preoperative GIK infusion increased myocardial glycogen and produced favorable effects, especially in preventing serious arrhythmias, it was not clear that the positive effects were mediated by an increased glycogen content.

Hemodynamic and Mechanical Parameters

It is useful to assess cardiac performance and compensatory interactions of a subject by observing both hemodynamic (pump related) and mechanical (muscle related) parameters of the left ventricle (Mason et al., 1976). Hemodynamic parameters have been successfully used to define the normal physiology and pathophysiology of the cardiovascular system. The progress in clinical cardiology and advances in cardiac surgery are to a large extent due to the application of hemodynamic principles (Yang et al., 1978). Cardiac physiologists are developing specific and sensitive mechanical indices of myocardial function. Myocardial mechanics defines the performance of the heart in terms of stresses, strains, and force-velocity relations in a fashion similar to the 3 element model of skeletal muscle (Maxwell Model) developed by A.V. Hill. E.H. Sonneblick concluded that the fundamental law of Hill for skeletal muscle also applies to myocardium; however, the study of myocardial mechanics has yet to provide specific indices of ventricular function (Yang et al., 1978).

Mechanical parameters are useful in assessing ventricular function when used in a comparative sense in the assessment of a single subject. For example, dp/dt, the rate of change of pressure within the chamber of the left ventricle, varies greatly from member to member within a species and is thus meaningless for intersubject evaluation. Nonetheless, dp/dt is a useful indicator of myocardial function when patients or animals act as their own controls in a study (Yang et al., 1978).

Left ventricular end-diastolic pressure

Left ventricular end-diastolic pressure (LVEDP) is a hemodynamic parameter that marks the instant at which the left ventricle is at its fullest. It is thus likely to reveal any abnormality in compliance due to hypertrophy or filling defects due to congestive heart failure (Yang et al., 1978). LVEDP is sometimes used as an indicator of left ventricular preload (Guyton, 1986). It is important to specify the preload of the left ventricle when evaluating the contractile properties of the left ventricle because the force of muscle contraction is affected by preload conditions as described by Starling's Law.

Mean left atrial pressure

In the absence of mitral valve obstruction or increased left atrial contraction with decreased left ventricular compliance, mean left atrial pressure (MLAP) closely approximates LVEDP (Mason et al., 1976).

dp/dt

Studies have indicated that the rate of rise of the ventricular pressure correlates well with the mechanical ability of the ventricular muscle. Peak dp/dt is thus used to compare the contractile ability of a heart in varied states (Guyton, 1986).

Left ventricular dp/dt is dependent upon preload, and in cases of aortic regurgitation where aortic end-diastolic pressure is greatly reduced, is also dependent upon afterload (Yang et al., 1978). Many indices are used in attempt to correct dp/dt for individual differences in preload and afterload. No one devised correction index is perfect, and careful consideration of the conditions of the study should be given in selecting a correction index.

Time to peak dp/dt is a corrected value that is useful for comparing changes of contractility in a subject as a result of mechanical interventions. This index is defined as the time interval from beginning of left ventricular contraction to peak dp/dt (Yang et al., 1978).

Peak dp/dt can also be divided by related quantities to correct for preload differences. When divided by left ventricular end-diastolic circumference, peak dp/dt can be adequately corrected for preload differences in an individual. However, this index suffers from the approximations inherent in determining the circumference of the ventricle. The results of dividing dp/dt by left ventricular end-diastolic volume correlates well with myocardial contractility except in cases of increased afterload or hypertrophy. In studies involving left ventricular hypertrophy, division by left ventricular end-diastolic pressure would be a better correction for the usually high dp/dt (Yang et al., 1978).

Peak dp/dt is calculated by passing the pressure waveform from the left ventricle through an analog or digital device (differentiator). A precision transducer with a frequency response greater than or equal to 30 hertz is required. Catheter-tip transducers are often employed for this task since this type of transducer usually has the required frequency response and its location reduces artifacts and damping of the pressure pulse (Yang et al., 1978).

Instantaneous flow measurements

Development of instantaneous flow measurement techniques to assess hemodynamics has lagged behind development of pressure measurement techniques. The measurement of pulsatile flow with acceptable accuracy became possible in the 1970s (McDonald, 1974). Two techniques for measuring instantaneous flow are electromagnetic and ultrasonic Doppler flowmetry. Both are capable of measuring pulsatile and mean flow. Electromagnetic flowmeters can display changes in flow that occur in less than 0.01 second to an accuracy level of +/- 15 % (Nihon Kohden Corporation, publication date unknown). Flow through a blood vessel is determined by the pressure difference between the two ends of the vessel and the impediment to flow due to vascular resistance (Guyton, 1986). Pulsatile flow is important in the evaluation of blood supply to cardiac muscle through the coronary circulation (Dinnar, 1981). Instantaneous flow measurements are often used to compare performance of coronary artery graft materials and techniques.

Segmental systolic shortening

Measurement of segmental systolic shortening (SS) is used to determine regional contractile performance of the ventricular muscle. Ultrasonic crystals, placed in the free wall of the ventricle, allow for instantaneous segmental length determinations. Segmental systolic shortening is then calculated as follows (Vinton-Johansen et al., 1986).

$$SS = [(EDL - ESL)/EDL] \times 100$$

Where EDL = end-diastolic length

ESL = end-systolic length

Results can be expressed as a percent of control SS to normalize data due to differences in crystal distances between subjects (Allen et al., 1986).

End-diastolic length is generally measured at the beginning of the left ventricular contraction. If a catheter tip transducer is used for determination of dp/dt, the time to measure EDL is indicated by the initial positive rise in the left ventricular dp/dt waveform. However, a phase shift inherent in fluid filled catheters requires that the measurement of end-diastolic length be taken 20 milliseconds prior to the initial positive rise in left ventricular dp/dt. Also, EDL generally coincides with the R-wave peak of electrocardiograph records (Kleinman et al., 1979).

End-systolic length is measured at the point of initial downstroke of negative dp/dt when catheter-tip transducers are used to determine dp/dt. To compensate for phase shift when fluid filled systems are used, end-diastolic length is measured at 20 milliseconds prior to initial downstroke of negative dp/dt. This point generally coincides with the midpoint of the T-wave of electrocardiograph records (Kleinman et al., 1979).

Data Acquisition Systems

Cardiothoracic experimental investigations require either a chart-recorder or a data acquisition system (DAS) to acquire hemodynamic and myocardial mechanical data. A DAS converts analog signals to digital form for storage, processing, display, and transmission (The engineering staff of Analog Devices, 1986). A DAS usually contains a computerized controller, an analog-to-digital converter, and an input multiplexer. Other devices such as circuits for sample and hold, input signal conditioning, and interfacing with special transducers may be included (Eckhardt et al., 1988). Personal computers (PCs) are becoming popular as the foundation for automated data acquisition and analysis. PCs offer flexibility, speed, attractive cost/performance, ease of operation, high resolution and precision. Also, an increasing selection of software allows for post acquisition computation, analysis, management, reduction, and presentation of data. In addition, the computer may be utilized for other tasks while not being used for data acquisition and analysis (National Instruments, 1989). Configuring a PC based system requires careful consideration and planning; the designer needs to evaluate the applications and select the appropriate computer, hardware, and software to meet the needs of the application (Judd and Phillips, 1986).

Analog-to-digital converters

A very important component of a computer based data acquisition system is the analog-to-digital converter (ADC). The ADC transforms analog data into discrete binary code or twos complement digital representation (Analog Devices, 1987). There are several different types of ADCs. Among other parameters, they can differ in resolution and speed.

Types of ADCs

Counter ADCs Counter ADCs use an up-counter and an internal digital-to-analog converter (DAC) to generate an analog output, one least significant bit at a time, to which the data signal voltage is compared. When the generated DAC voltage output equals the data signal level, a comparator stops the counter. The final digital output is equal to the final count. Counter ADCs are not very fast and suffer from inconsistent conversion times since the conversion time varies with the level of the input signal (Sood, 1988).

Tracking or servo ADCs Tracking or servo ADCs are a variation of the counter ADC. Tracking ADCs use an up/down counter to allow the DAC voltage to continuously follow or "track" the input data signal. By stopping the counter the tracking ADC gives a digital output which represents the signal level at that time (Sood, 1988). Tracking ADCs offer high speed in track; however, they cannot follow large data signal changes and are susceptible to noise (Analog Devices, 1987).

Integrating ADCs Integrating ADCs approximate integration by taking the average of the data signal voltage over a set period of time. The averaged signal is put into a capacitor which is discharged at the end of the period of integration. The ratio of discharge time to integration time is used to quantify the digital solution (Real Time Devices, 1988). Integrating ADCs offer excellent noise rejection as the integrating converter averages out the effects of noise. This type of converter is very accurate and inexpensive. On the other hand, integrating ADCs are limited by low speed (Sood, 1988).

Voltage-to-frequency ADCs Voltage-to-frequency ADCs use a precision voltage-to-frequency converter (VFC) to change the data signal voltage into a train of pulses with a frequency proportional to the data signal voltage. A digital representation of the data signal is then generated by a counter which counts the number of pulses over a set period of time. This type of converter is characterized by slow speeds with good noise immunity (Sood, 1988).

Parallel or flash ADCs Parallel or flash ADCs are used in applications that require very high speed conversion. Flash converters operate by simultaneously comparing the input voltage signal to the threshold values of several comparators. The comparators are arranged in parallel with the threshold values biased one least significant bit (LSB) apart. For a zero input voltage signal, all of the comparators are in the off state (The engineering staff of Analog Devices, 1986). As the input voltage signal increases above zero, the comparators, which are biased at a level below the input signal, change state to on. A fast encoder then converts the comparator states into an output code (Sood, 1988). Flash converters are evolving rapidly; current models are capable of conversion rates of several megahertz.

Successive-approximation ADCs Successive-approximation ADCs are the most commonly used ADC since this type of converter offers a good combination of speed and high resolution. It is possible to obtain sixteen bits of resolution at a sampling rate of 1 megahertz (Analog Devices, 1987). This type of converter uses an internal DAC. However, unlike the counter ADC, the successive-approximation DAC is able to match the level of the input voltage signal very quickly. As a result, the conversion time of a successive-approximation ADC is very fast. Also, the time of conversion is not dependent upon the level of the input voltage--a major disadvantage found with the counter ADC (Sood, 1988).

The successive-approximation internal DAC approximates the input signal level with a binary code which is put into a register within the device. A comparator compares the approximation with the input signal voltage and commands the addition or subtraction of bits of code, into the register, until the best approximation is obtained. This approximation within the successive-approximation register is then available to the computer as the final digital word (Sood, 1988).

Resolution

The resolution of an ADC is the smallest increment of measurement that the ADC can recognize. This measurement is usually a voltage represented by the LSB of the converter word (Analog Devices, 1987). Resolution is usually listed in ADC performance specifications as the number of bits used to quantize the input signal. More bits means more recognizable divisions into which the ADC input signal range can be divided (National Instruments, 1989). An 8 bit ADC that has an input signal range of 10 volts will have a resolution or LSB voltage as calculated below (Sood, 1988).

$$\text{Resolution} = \text{LSB Voltage} = 10\ \text{V}/[(2)^8] = 0.039\ \text{V}$$

Some transducers have very wide dynamic ranges. Dynamic range is the difference between the smallest detectable signal level and the largest full scale signal level (Analog Devices, 1987). A high resolution ADC (16 bit) may be necessary to utilize the accuracy of a transducer with a wide dynamic range.

In some applications, an instrumentation amplifier (IA) may be used to amplify the transducer voltage to enhance the sensitivity of a medium resolution ADC. The amplification offered by an IA allows smaller input voltage ranges to be brought directly to the converter. In such a case, the input voltage range is made smaller by a factor corresponding to IA gain and the LSB voltage or resolution will thus be increased (Analog Devices, 1987). Using the previous calculation for an 8 bit ADC, the addition of an IA with a gain factor of 16X, produces the following calculated resolution.

$$\text{Resolution} = 10\ \text{V}/[(2)^{8} \times 16] = 0.0024\ \text{Volts}$$

Notice that the resolution gained by using an IA set at 16X, with an 8 bit ADC, is theoretically equal to the resolution acquired by a 12 bit ADC, with no IA gain, as shown below.

$$\text{Resolution} = 10\ \text{V}/[(2)^{12}] = 0.0024\ \text{Volts}$$

Speed

Speed of A/D conversion is usually listed as throughput rate. Throughput rate is the inverse of the acquisition time. Throughput rate of the ADC is an inconclusive figure as other acquisition components influence overall speed. As a system specification, throughput rate takes into consideration multiplexer (MUX) settling time, IA settling time, sample-and-hold amplifier (SHA) aperture and settling time, A/D conversion time, and time of data transfer. It should be noted that if an adjustable gain IA is part of the system, then the IA gain setting will affect the settling time of the IA and thus affect overall conversion speed. Higher gain corresponds to a higher settling time and thus a slower conversion rate. Measurements have shown that certain ADC board throughput rates, when compared to throughput rates at unity gain, decrease by as much as 65% when using a gain factor of 500X (Analog Devices, 1987). As an approximation, MUX and IA settling account for 20%, SHA settling accounts for 10%, A/D conversion accounts for 55%, and data transfer accounts for 15% of total acquisition time (Analog Devices, 1987).

Frequently, throughput rates are given as an ADC board level specification. The performance will vary with different types of computers. In single ADC multichannel systems, board throughput rate is divided by the number of sampling channels to determine per channel throughput at the board level (Metrabyte Corporation, 1989).

Max. Throughput/Channel = Max. Board Throughput/# Channels

An 8 channel board rated for 10 kHz throughput will thus have a maximum sampling rate of 1.25 kHz/channel.

Digitizing multiple channels

Two approaches are generally used to simultaneously digitize more than one analog input signal: analog multiplexing and parallel conversion (Sood, 1988).

Multiplexing An analog MUX is a set of switches that route several channels of data to a single ADC. The MUX is commanded by control logic to connect one channel at a selected time to a SHA for A/D conversion. Each channel can then be sequentially converted in a make and break fashion without experiencing time-skew problems. Most analog multiplexers use semiconductor JFET or CMOS switches and come in a variety of configurations: 2,4,8,16,32,64...256 single and/or differential channels. Important multiplexer specifications are on-resistance, leakage currents, transfer accuracy, crosstalk, and settling time (Sood, 1988).

<u>Parallel conversion</u> With the recent drop in price of ADCs, parallel conversion has become increasingly popular (The engineering staff of Analog Devices, 1986). In parallel conversion a separate ADC is used for each input channel. This offers many advantages over analog multiplexing. Since each ADC is converting just one channel, extremely fast A/D conversion is possible. At medium sampling rates, a cost savings may be realized since slower and less expensive ADCs can be used and SHAs may be eliminated (The engineering staff of Analog Devices, 1986). The parallel conversion scheme allows the ADCs to be placed near the transducers, and data can then be shipped to the DAS in digital form. This is useful in electrically noisy environments (Sood, 1988).

Bus structures used by microprocessors favor the use of digital multiplexing. Digital logic circuits can be used to exercise judgement on sending data to the host computer. This can decrease redundant sampling and optimize conversion and storage processes. This is not possible with an analog multiplexed system until after the signals have been digitized. This requires extra time and energy (The engineering staff of Analog Devices, 1986).

<u>Data transfer</u> Once data are digitized by the ADC it can be transferred to the computer in a number of ways. High speed data acquisition over a long time interval may require use of direct memory access (DMA) or ADC on-board memory.

Direct memory access Direct memory access (DMA) provides fast transfer of data by temporarily suppressing the central processing unit (CPU) of the computer and routing the digitized data directly to the computer's random access memory (RAM) (Kim, 1981). A DMA controller takes command of the computer's bus during data transfer and generates required addresses and control commands. At the end of this interval, the bus control is turned over to the microprocessor (Sood, 1988). Most PCs are restricted to DMA segments of 64 kilobytes. To contiguously store more than 64 kilobytes requires a first-in first-out (FIFO) buffer to temporarily hold data while the computer switches to another 64 kilobyte segment (Analog Devices, 1987).

On-card RAM memory Another way of circumventing the 64 kilobyte DMA restriction is to use on-card RAM memory to store acquired data. This scheme does not use DMA at all. RAM memory, installed on the ADC board, holds digitized data until the PC can store the data. This results in faster acquisition times as data and address bus access times are eliminated and the buses are thus free for the CPU to further speed data transfer (Analog Devices, 1987).

Sample-and-hold-amplifiers

A SHA is used to hold an input signal at a constant level while the ADC converts the signal to digital form (Analog

Devices, 1987). A SHA reduces uncertainty error in the analog-to-digital (A/D) output when the input signal change is fast compared to the A/D conversion time. As a result, a SHA helps increase overall speed of accurate conversion. SHAs are also used in some multichannel systems to hold the input signal of one channel for conversion while another channel's input is being digitized (Sood, 1988).

A SHA consists of a storage capacitor, input and output buffer amplifiers, and a control switch which is sometimes called a track/sample switch. During the sample period, the control switch connects the input signal voltage to the storage capacitor which is rapidly charged to the input signal level (Analog Devices, 1987). At the end of this acquisition time, the control switch opens and the capacitor then holds the signal until it is converted. There are two time delays to be considered during the beginning of this hold period. First, there is a time delay between the command to hold and the opening of the control switch. In systems that use SHAs this aperture delay determines the maximum frequency of input signal allowable if full conversion accuracy is to be maintained. Aperture times are usually of the order of a few nanoseconds (Sood, 1988). Second, after the control switch opens, a settling time must pass before the input voltage stabilizes enough for faithful conversion to take place. The input voltage cannot stay in hold for too long since leakage or capacitive bias currents will cause the signal level to drop off or droop. Acquisition time, aperture time, settling time, and droop rate are all specifications to consider when examining SHAs.

Signal Conditioning and Transducer Interfacing

It is sometimes necessary to condition the input voltage signal in order to view and acquire data in the desired form. Common signal conditioning includes amplitude adjustment (usually amplification), filtering, differentiation, integration, peak and valley detection, and isolation. Excitation voltages may be necessary to drive active transducers, and reference voltages may be required for standardization and/or calibration of the DAS.

Instrumentation amplifiers

IAs are designed to supply high input impedance/common mode rejection, low ambient temperature induced offset and drift, low nonlinearity, and stable accurate gain (Analog Devices, 1987). If input signals span only half of the total ADC input range then any converter errors are effectively doubled (Sood, 1988); therefore, IAs may be used to increase transducer output voltage to best utilize the ADCs input range and enhance the ADC's sensitivity.

IAs may have software controlled gain/offset and may be supplied on the A/D conversion board. Software programmable gain is a convenience that allows for keyboard control of individual channel's gain/offset.

IAs alone may offer little signal isolation or noise rejection. However, amplification of small transducer signals, at the site of transduction, can sometimes be useful in electrically noisy environments. For example, induced noise of 10 microvolts added to a transducer signal of 10 microvolts causes 100% error (Analog Devices, 1987). Amplification of the signal at the transduction site by 100X can reduce this induced noise error by as much as 100X.

Isolation amplifiers

Isolation amplifiers isolate transducer signals from the computer. For example, the Analog Devices Model 248J high performance transformer isolation amplifier offers common mode voltage (CMV) isolation of +/- 5000 V peak value delivered over a 10 millisecond pulse and continuous CMV isolation of +/- 2500 VDC (Analog Devices, 1985). Isolation serves to protect the computer and operator from transient voltage-spikes picked up by transducer-leads. In clinical or physiological investigations in which the subject may be electrically susceptible, isolation amplifiers protect the subject from transient or computer generated electrical potentials that could cause microshock. "Optical isolation or transformer isolation of the electrical leads connected to the patient is probably the best way to protect patients from most macroshock and microshock hazards" (Olson, 1978).

Filters

Filters are sometimes necessary to eliminate unwanted noise or to extract a particular feature of an input signal. Filtering may be accomplished by either analog or digital techniques. It should be noted that noise elimination filters are not a justifiable substitute for proper wiring, layout, and shielding techniques. However, filters may prove useful in adjunct to a properly engineered DAS (Analog Devices, 1981). Care should be taken when filtering input signals to insure that important frequency components of the input signal are not being inadvertently lost.

Analog filtering   Analog filters process the input signal in a continuous fashion while the signal is still in analog form. There are four types of filters: low-pass, high-pass, band-pass, and band-reject or notch (Coughlin and Driscoll, 1987). Most present day analog filters are active filters constructed of operational amplifiers and resistive/capacitive networks.

Analog filters can be constructed by someone who does not have an extensive electrical engineering background. Transfer functions, which mathematically express the relationship between input signal and output signal, are available in algebraic approximation equations. These equations can be found in integrated circuit data books which most operational amplifier manufacturers will supply. By changing resistive and capacitive values, a variety of filter response characteristics can be achieved from one circuit. Filter construction often requires a great deal of time, and a bipolar power supply, oscilloscope, and function generator are required to construct and test most filters. Alternatively, complete and tested active filters can be purchased from several manufacturers.

Digital filtering Digital filters are either digital electronic circuits or computer programs that process digitized samples of the input signal to achieve an output similar to that produced by analog filtering. Digital filters operate on a sequence of numbers, each of which corresponds to a specific instant of time. Each instant of time corresponds to an A/D conversion sampling point (Baharestani and Tompkins, 1981). The necessary operations of any digital filter are (1) storage, (2) multiplication by constants, and (3) addition.

J.F. Kaiser defined a digital filter as follows: "The term *digital filter* refers to the computational process or algorithm by which a sampled signal or sequence of numbers (acting as an input) is transformed into a second sequence of numbers termed the output signal" (Strum and Kirk, 1988). A PC may be used to implement a real time digital filter if its microprocessor is fast enough to perform the necessary filter operations and produce a number at its output for each number that it receives from the A/D converter (Baharestani and Tompkins, 1981). Real time digital filtering of a single input signal is relatively easy to implement if the filter algorithm is not too complex. For example, digital low-pass filtering in real time can be accomplished on most PCs. However, with current techniques, it is difficult to continuously display both a filtered output with its input signal in time synchrony at the time of the event. Also, when more than one digital filter is required in a multichannel recording situation, real time filtering with a standard PC is usually precluded (Baharestani and Tompkins, 1981). In cases such as this, digital filtering is usually done in post acquisition (after input data has been recorded), or an analog filter is used.

There are two basic digital filter types. There are nonrecursive and recursive filters. They differ from one another in the structure of the filter's transfer function. The transfer function of a digital filter is a polynomial expression that relates the filter input to the filter output. The output of a recursive filter system depends upon previous values of the output as well as the current values of the input. In nonrecursive systems, the previous output values are not included in the solution (Strum and Kirk, 1988). Nonrecursive filters are known to be consistently stable, completely linear in phase response, easy to design, and stable in response to input disturbances such as switching transients. However, nonrecursive filters, unlike recursive filters, cannot produce a sharp cutoff frequency with a fast rolloff (Baharestani and Tompkins, 1981).

Digital filters offer many advantages over analog filters. They have both high noise immunity and accuracy. Digital filter accuracy corresponds to the roundoff error which can be made small during filter software development. Digital filters are cheaper than analog filters. It is also easy to change digital filter characteristics. With appropriate software, filter characteristics may be changed on command by pressing a key on the PC keyboard. Digital filters are not affected by temperature changes, component aging, or power supply fluctuations. As a result, digital filters offer excellent repeatability (Baharestani and Tompkins, 1981).

Anti-aliasing filters Aliasing is a phenomenon that occurs when the sampling frequency of the DAS is less than twice the highest frequency of the signal being digitized. There will always be some aliasing present whenever analog signals are converted to digital signals. This is due to high-frequency components present in the analog signal, whether they are harmonics of the signal or contamination from noise (Williams, 1986). Aliasing errors occur when significant information contained in frequencies higher than the resolution allowed by the sampling rate appear mixed with lower frequency components. The result is signal distortion. The effects of aliasing can be significantly reduced by use of an analog low-pass filter that attenuates the unnecessary high-frequency components and bandlimits the analog signal to at least one half the sampling frequency before it is digitized. An analysis of the frequency components of the signal of interest is necessary before designating the filter transfer function. This is done to insure that important components of the signal are not attenuated and lost, thus creating an area of omission (Monzon and Tompkins, 1988). Anti-aliasing cannot be accomplished with any digital filtering process (Baharestani and Tompkins, 1981).

Analog averaging techniques The mean of an input signal is the zero harmonic or direct-current (DC) level of the signal as determined by Fourier approximation (Rubin, 1987). This signal may be determined by using a low-pass filter or by using a process of integration. Integration is commonly used to determine a mean signal over a given time period. Low-pass filters work best when observing the mean signal in a continuous manner (Rubin, 1987).

An ideal analog low-pass filter outputs a constant voltage from DC to a cutoff frequency, f_c. In reality, f_c is usually at the frequency where gain is reduced to 0.707 of its low-frequency value (Coughlin and Driscoll, 1987). As the input signal frequency increases above f_c, the output signal is attenuated. The rate of attenuation, or rolloff, is often related in terms of decible/decade (dB/decade) where a decade equals a frequency factor of 10X. Attenuation is a circuit closed-loop gain figure approximated as follows.

Gain in dB = $20 \log_{10}[V_o/V_i]$

Where V_o = filter output voltage

V_i = filter input voltage

For example, if filter gain decreases by 20 dB/decade, then gain is divided by a factor of 10 as the input frequency increases by a factor of 10 (Coughlin and Driscoll, 1987). Analog low-pass filters may be cascaded to enhance rate of attenuation. The overall gain of cascaded filters equals the product obtained by multiplying together each filter's gain. In the complex domain, low-pass filters's responses have many poles which are equivalent to the order of the filter. This in turn corresponds to the number of storage elements found in the filter (Analog Devices, 1981). There are many types of low-pass filters such as Butterworth, Bessel, Chebyshev, and

Paynter; the performance of these varies in response to a choice of time constants, natural frequencies and damping of filter elements (Analog Devices, 1981).

<u>Butterworth low-pass filters</u> Butterworth filters work well in low-pass applications as they have a flat closed-loop gain of close to 1X in the pass band. Butterworth filters are not designed to keep a constant phase angle at the cutoff frequency. The phase angle at f_c increases with each increase in attenuation. Filter output lags behind input signal by 45° at f_c in a -20 dB/decade Butterworth filter. For each increase of -20 dB/decade attenuation, the phase angle, or output delay, increases by approximately 45° (Coughlin and Driscoll, 1987). This phase angle will not be a problem in applications in which mean signal is being viewed over a long time.

<u>Moving average filter</u> The moving average filter is a much used nonrecursive design which approximates an analog low-pass filter. The designer selects a number of data points to be successively averaged by the filter. The larger the number of points, the greater the smoothing effect produced. Also, more points means more time required for the filter to calculate a response. For example, in a 200 point moving average filter, the first 200 points that enter the filter will be averaged producing one output number. As the next sequence of numbers enters the filter, the filter will average

only the most recent 200 points, again producing an averaged output number. This process is successively repeated for all data points resulting in a cumulative output that is fit to the averaged points (Strum and Kirk, 1988).

The Hanning filter is a form of nonrecursive moving average filter that produces an output data point that is the scaled average of three successive input points. The center point is weighted twice as heavily as its two adjacent points (Baharestani and Tompkins, 1981). This filter produces a characteristic passband of a low-pass filter with a gain of one at DC and a rolloff to zero when the input frequency equals one half of the sampling frequency.

Least-squares polynomial fitting A least-squares polynomial fitting low-pass nonrecursive filter approximates sets of input data points by a parabolic polynomial. Any odd number of points may be selected to be fit by a parabola. The process repeats for each new input data point. As an example, if a 7 point filter is chosen, a parabola will be fit by the algorithm to the first 7 points that enter the filter. When data point 8 enters the filter, a new parabola will be fit to this point and the previous 6 points. This process repeats for each new input point and a 7 point parabola is thus fit for every new input point. The output of the filter is then the center points of all of the calculated parabolas. The greater the number of points chosen for the filter, the greater the rate of rolloff. However as the number of points increases, filter processing time increases as well. This technique also approximates the effect of an analog low-pass filter (Baharestani and Tompkins, 1981).

Differentiation

A differentiator is a device that approximates the mathematical process of differentiation. Differentiators respond to changes in the input signal level and not to the absolute level of the input signal. Differentiation of time dependent signals characterizes the rate of change with respect to time. Maximum, minimum, or zero changes can be indicated by differentiator output signals. In frequency terms, a differentiator produces a rising output in response to increasing signal frequency. Although an ideal differentiator is not limited by frequency response, a real differentiator will have a cutoff frequency at which the output signal fails to increase with frequency (Rubin, 1987). Approximation of differentiation may be accomplished by either analog or digital techniques.

Analog differentiation Most analog differentiators are active, utilizing operational amplifiers with RC networks to best approximate differentiation by an analog technique. Analog differentiators offer real time processing of the input signal at a moderate cost. However, high-frequency noise and instability problems are inherent in analog differentiators (Woodburn and Tompkins, 1981).

Digital differentiation It is recognized that differentiation is not defined for digital data (Williams, 1986). However, differentiation of input signals that are continuous can be closely approximated by a number of techniques such as secant-line approximation, forward or backward-difference approximation, and polynomial modeling. These methods usually involve trade-offs of increased accuracy and resolution for increased computer memory and computation time. All algorithms contain margins of error that decrease as the interval between input points decreases (Woodburn and Tompkins, 1981).

Digital filters cannot indefinitely increase output signal level in response to increasing frequency of input signal. The amplitude response of most digital differentiators is limited to 1/2 the sampling rate used to acquire the input signal. At this signal frequency the output signal level drops to zero (Woodburn and Tompkins, 1981). It is therefore important to pay close attention to the bandwidth characteristics of both the input signal and the DAS to insure adequate processing.

In general, digital systems of differentiation offer the stability and repeatability inherent in a numerical algorithm, noise reduction through smoothing, and software controlled changeable parameters (Woodburn and Tompkins, 1981).

Polynomial modeling is a technique in which a best-fit polynomial expression is derived to model the input signal. The slope of the polynomial expression is determined and then used to compute an output signal from the continuous data signal that has been digitized. The resulting derivative expression will also be a polynomial. The order of the derivative polynomial expression varies with the number of data points used in the model (Williams, 1986). The value of the approximation of the derivative improves when more than 5 points are used to fit the polynomial expression. This type of derivative algorithm, which is nonrecursive, offers a smoothing characteristic that helps attenuate high-frequency noise commonly seen in other derivative algorithms (Woodburn and Tompkins, 1981). Also, gain is not a function of sampling period as in other algorithms. Forward or backward difference approximations are special types of polynomially derived derivatives (Williams, 1986).

SELECTING A DAS

Selection and configuration of a DAS requires careful consideration of applications in which the DAS is to be used. An evaluation of data to be recorded and the measuring devices used to present the data signals to the DAS will help determine basic requirements such as the following:

1) Throughput rate
2) Resolution
3) Available input/output channels
4) Type of input channels available
   (single ended and/or differential input channels)
5) Input voltage range
6) Gain/offset per channel

It is also necessary to consider the studies in which the DAS may be used to determine other DAS requirements such as:

1) Type of display format
   (real time and/or post acquisition graphical and/or numerical displays)
2) Data storage format
3) Playback and analysis capabilities

4) Data conditioning capabilities

5) Waveform reproduction

Other factors which must be considered before selecting a DAS are as follows:

1) Computer/software compatibility

2) Supported printers/plotters

3) Expansion capabilities of the DAS

4) Time available to develop and test DAS

5) Cost of DAS including operation/maintenance

Signal Analysis

The signals of interest need to be sampled fast enough to faithfully represent the data as seen by the DAS. The sampling theorem states that if a bandwidth limited signal contains no frequency components higher than value f_x, then the original signal can be reclaimed completely without any distortion if it has been sampled at a rate at least two times f_x (Yanikowski, 1981). Sampling rates greater than $2f_x$ can increase resolution but require a larger storage space. Also, the price of the ADC increases with its throughput capabilities. Most designers select sampling rates that are at least 20% greater than the minimum required by the sampling theorem and strive for higher rates, approaching the limits imposed by recorder space or DAS memory and conversion capabilities (Williams, 1986).

The bandwidth of biological signals spans from DC to 20,000 Hz. However, most physiological signals are at the lower end of this spectrum. Care must be taken when reading bandwidth values from references. Many sources do not state critical information such as species or state of subjects from which bandwidth data was derived. It is known that the cardiovascular fundamental frequency and bandwidth requirements for studying an exercising animal are 2 to 3 times greater than for studying the same animal at rest (Rubin, 1987). The bandwidth of cardiovascular parameters of a resting dog are usually greater than that of a resting human (Yang et al., 1978).

In addition, different forms of analysis or processing planned for a signal may mandate different bandwidth requirements. For example, dp/dt requires a different bandwidth from that required for the ventricular pressure signal.

Blood pressure bandwidth

Shirer (1962) gives canine cardiovascular bandwidth requirements of 30 Hz to 100 Hz in order to insure faithful reproduction of the pulsatile waveform. This covers the mean (DC or zero harmonic) component, the fundamental frequency (the repetition frequency of the wave itself) plus 6 to 30 harmonics for canines with a heart rate from 1.5 to 5 beats/second. Olson (1978) lists a bandwidth of DC to 50 Hz for direct arterial measurement, however, heart rate and species are not given. In pressure measurements the zero harmonic (the constant term) is the largest component. Higher frequency terms are primarily necessary to describe sharp fluctuations in the waveform such as the dicrotic notch seen in the aortic pressure wave or the a, c, and v atrial pressure waves (Caro et al., 1978). It is generally accepted that a bandwidth of 10 times the fundamental frequency of the waveform will faithfully reproduce the waveform. It is not uncommon for adult dogs to have a heart rate up to 160 beats/minute (BPM). Younger dogs can exhibit rates up to 220 BPM. However, it was anticipated that during the study involving GIK, the anesthetic would keep heart rates near 120 BPM. A heart rate of 120 BPM or 2 Hz requires a bandwidth of 20 Hz for faithful reproduction (Yang et al., 1978).

dp/dt

The first derivative of a blood pressure waveform produces frequencies higher than those of the waveform itself. Fourier frequency analysis of the derivative signal, but not of the raw waveform, is thus required to accurately determine the bandwidth requirements. This type of information is rarely available (Rubin, 1987). Peura (1978) states that for ventricular waveforms that will be differentiated the amplitude to frequency response of a catheter-manometer system should remain flat to within 5% up to the 20th harmonic of the fundamental of the raw waveform. Yang et al. (1978) give 30 Hz or greater as an adequate natural frequency of a transducer system for determination of dp/dt.

Flow rate

McDonald (1974) has stated in relation to blood flow that no significant propagated waves exist above 30 Hz and that those found above 20 Hz are small. Olson (1978) specifies DC to 20 Hz for a blood flow range from 1 to 300 ml/second. In aortic flow rate measurements, the zero harmonic is smaller than the fundamental frequency with representation from higher harmonics necessary to lend resolution during the rapid rise of flow in early systole and possible sharp fluctuations of flow reversal that occur at the end of systole. Further away from the heart where the pressure waveform is smoother, fewer harmonics are required (Caro et al., 1978).

Electrocardiograph

Olson (1978) specifies a conservative bandwidth of 0.01 to 250 Hz for EKG signals. Jacobson and Webster (1977) state that an upper frequency limit of at least 100 Hz is necessary for faithful EKG reproduction. Poor high-frequency response can limit the amplitude of EKG components such as the QRS complex. The low-frequency limit should extend down to 0.05 Hz which corresponds to a time constant of 3 seconds. A shorter time constant will distort the P and T components of the EKG, as well as the ST segment, decreasing the worth of the EKG for diagnostic or timing determinations. The 0.05 to 100 Hz EKG bandwidth is also a specification of the American Heart Association to ensure clinical accuracy (Neuman, 1978).

Measuring Devices

The first set of experiments, called the GIK study required the simultaneous use of two active pressure transducers, one bipolar EKG lead, and one ultrasonic device to measure distance between a pair of transducers implanted in the heart. Each pressure transducer would be required to simultaneously supply more than one waveform to the DAS for analysis. One transducer supplied pulsatile and MLAP to the DAS. The second transducer supplied pulsatile left ventricular pressure (LVP) and dp/dt from the LVP waveform. In addition three other studies required the simultaneous use of two electromagnetic flowmeters, a bipolar EKG lead, two pressure displays, and the sonomicrometer.

<u>Pressure transducers</u>

Millar MPC-500 Mikro-Tip catheter pressure transducers were selected because of their capability to produce high fidelity pressure waveforms. By eliminating the hydraulic connection between pressure source and transducer, time delay of pressure pulse transmission is eliminated and frequency response is enhanced. These two considerations are especially important when the transducer is to be used to determine dp/dt. Also, hydrostatic pressure errors and catheter whip artefacts are eliminated when there is no hydraulic connection in the transducer system. This model transducer is considered to be somewhat fragile and expensive; it is without warranty after receipt. Special care is required in cleaning and handling of the transducer to insure that it functions properly and is not damaged.

Each Millar pressure transducer requires a control unit for interfacing to a chart-recorder, monitor, or DAS. Millar model TCB-500 control units were selected for use since they supply bridge excitation, balance, and calibration control for each transducer. These battery powered control units also supply amplification giving a nominal output signal of 0.2 V/100 mmHg. Battery power offers some microshock protection for the subject. The unit's internal calibration signal outputs electronic zero as well as 20 mmHg and 100 mmHg electronic calibration signals. Having both 20 and 100 mmHg calibration levels allows calibration output to the DAS in two ranges: venous and arterial.

Sonomicrometer

Triton Technology's Sonomicrometer 120 ultrasonic distance measuring device was selected for use in the GIK study as the device has been validated and used extensively in animal models to assess global and regional myocardial dimensions (Hill et al., 1978). This device measures transit time between two piezoelectric transducers placed in the myocardium and outputs an analog voltage proportional to the distance between the two crystals. Up to four distances, in close proximity, can be measured simultaneously without interference. A calibration signal is supplied by switching to an internal clock which outputs a stair-step DC voltage in one millimeter increments. Output voltage offset and gain are both adjustable on each channel making the unit compatible with most chart recorders, monitors, and DASs.

Each of the four channels is supplied with an EKG circuit which allows unipolar or bipolar lead electrogram measurements using the piezoelectric transducers and a ground-wire attached to the subject. Ground reference for the EKG circuit is not connected directly to chassis ground but is attached to the output of an amplifier. Common-mode voltage is sensed by the amplifier, inverted, and fed back to the subject through the ground-wire, driving common mode voltage to a low value. This driven system eliminates noise and offers some isolation protection to the subject in the event of an abnormally high voltage potential between the subject and ground. An internal voltage pulse is also provided for calibration of a chart recorder or DAS.

Datascope 861 portable physiology scope (EKG monitor)

This device is a clinical monitor that displays two pressures and an EKG signal. One of two available auxiliary single ended outputs could be used to display and record EKG on the DAS. However, the low end frequency response of the EKG channel limits the accuracy of this device. The EKG channel has front end transformer isolation circuitry for patient safety. Leads I, II, III, aVL, aVR, aVF and V are available.

Flowmetry

Two electromagnetic flowmeters and a selection of precalibrated probes were loaned by Nihon Kohden Corporation of Japan. The MFV-3100 and MFV-3200 flowmeters were designed to work synchronously with two probes placed more than three centimeters apart. Simultaneous output of the pulsatile blood-flow waveform, mean blood-flow with time constants of one and three milliseconds, and stroke volume are possible with the MFV-3200. Stroke volume output requires an EKG signal input to the MFV-3200. The MFV-3100 can output either pulsatile or mean flow at one time. With the MFV-3200 coupled to the MFV-3100 in synchronous operation, a total of five output signals is possible using two probes. However, the experiments do not call for more than two outputs from two probes at any one time. Both units have digital LED displays; the MFV-3100 is capable of displaying a digital output of mean flow and the MFV-3200 is capable of displaying digital mean flow or stroke volume.

The pre-calibrated probes are each supplied with a "Cal-pack" that contains specific calibration data determined by physiological saline flow trials conducted by Nihon Kohden. Zero and calibration voltages can be output to a DAS from either flowmeter by using Cal-pack information. Output voltage gain is adjustable at each flowmeter making it easy to set the output voltage for each individual probe to optimize the full scale channel input range of the DAS. Also, each unit has isolated input and excitation circuitry for subject microshock protection and instrument protection against defibrillators and electrocautery (Nihon Kohden Corporation, publication date unknown).

DAS Requirements Determined by Signal and Instrument Specifications

Throughput rate

The range of reported EKG bandwidths was 0.01 to 250 Hz. Consequently, it appeared that EKG signals would require the greatest single channel throughput rate. For this bandwidth, a DAS channel with a 500 Hz throughput rate would be required for faithful reproduction of EKG. Moreover, previous experience demonstrated that sampling rates up to 10X the bandwidth might be necessary to achieve operational resolution of a digitized signal. This put the estimated necessary EKG throughput at 2500 Hz. As most DAS systems under consideration used a single ADC and were limited to an equal throughput rate per channel, the overall minimum throughput for seven channels would be 17,500 Hz. The available 8 channel systems would need an overall throughput of 20,000 Hz and 16 channel systems would need have 40,000 Hz capabilities.

Resolution

Eight bits of resolution would likely be adequate for the immediate studies. Operational or instrumentation amplifiers could be used to prescale input signals to enhance sensitivity of an eight bit converter as described in the resolution section of the literatue review.

Available input/output channels

At least seven input channels would be required. As some inputs would be differential and others single ended, a DAS with the ability to receive both types of inputs was preferred. However, operational amplifiers could be used to convert differential signals to single ended to match a DAS that could not be programmed for both types of input.

Input voltage range

Most available DASs have input ranges from either 0 to 5 V or 0 to 10 V in single ended input systems. In double ended input systems, +/- 2.5 V and +/- 5.0 V ranges are common. Any of these would work although amplification of the input signals to best utilize the DAS's input range would be necessary. The flowmeters and sonomicrometer have adjustable signal output gain. However, gain and possibly offset would be required for pressure and EKG signals. Although gain/offset conditioning would not have to be supplied by the DAS, it would be a convenient feature.

Type of display format/data storage format

At least 7 channels of real time display, with waveform scaling in familiar units, would be necessary in order to analyze the data from and control the experiments. For example, real time MLAP was necessary to control the blood volume infused after weaning a dog from the heart-lung bypass machine.

Simultaneous recording to disk with keystroke on/off control to select the intervals recorded would also be required. It was not known how large each file would be for any one experiment, however, files would have to be stored on floppy disk or tape for later access and as a backup precaution against hard disk failure.

Playback, analysis, and data conditioning capabilities

Playback of waveforms that were scaled in familiar units would be necessary to analyze and extract data. The ability to develop or access programs that could digitally condition the data would be required. In addition, calculations such as dp/dt are done better by digital means. Moving average algorithms might be necessary to extract usable data from a noisy signal that could not be conditioned by analog means.

Waveform reproduction

The DAS would need to support a printer to produce documentation of waveforms.

AT-Codas DAS

The DAS selected was the AT-Codas hardware and software package including optional Advanced Codas computational software and high performance input/output (I/O) module from Dataq Instruments, Inc. This ADC offers 12 bits of resolution (1 part in 4096) capability for 16 single ended or 8 differential analog input signals in any combination. Programmable system throughput to 50 kHz is possible. The high performance I/O module buffers and multiplexes the analog input channels and can supply channel independent gains in steps of 1, 2, 5, 10, 50, 100, 500, and 1000 times the input signals. Each input channel has full range offset control (+/- 5 V). These amplifier functions are software selectable by simple keystroke commands through Codas software. In addition, digital inputs and outputs (8 each), as well as 1 analog output, are available but are not supported by Codas software. An optional Microsoft C development software package can be used to access these lines for general purpose applications.

During real time, 9 display formats support programmable overlapping and nonoverlapping channels in a smooth scroll waveform. From 1 to 16 channels can be displayed in these 9 formats at a maximum of 2000 pixels/second/channel. In post acquisition mode, 10 displays are available. Data can be compressed, scrolled at various speeds, and stopped with single keystroke commands. Waveform display scaling and offsetting is available on each channel to expand or reposition the display for convenience.

Up to 29 channels may be displayed in post acquisition. This includes up to 16 recorded channels plus calculated channels generated by the optional Advanced Codas software. A cursor that spans full screen, in conjunction with a channel by channel alphanumeric scaled unit display, makes it easy to extract data points at any time during post acquisition.

Data storage is by DMA directly to disk while maintaining a real time display. The maximum storage rate is 50 KHz to a hard disk.

Advanced Codas software

Optional Advanced Codas software was purchased to supply digital signal conditioning. During post acquisition, Advanced Codas software can calculate and generate scaled waveforms from acquired data. Functions available include waveform rectification, integration, differentiation, moving average filter, and cycle by cycle peak capture. This software allows functions to be used individually or in any combination with automatic calibration of calculated units. Calculated channels can be aligned in time and displayed simultaneously with input data channels.

On-board signal processor

A TMS 32010 on-board signal processor uses a modified Harvard architecture that is optimized for high performance number processing. This form of architecture, which supports separate program and data memory spaces, allows for overlap of instructions (execution of one instruction overlaps the fetch cycle of the next instruction) as well as parallel fetch of instructions and data. Accepting and delivering 16 bit data signals but operating internally on 32 bits, this device insures precision and speed up to 5 million instructions per second (Mips) (McDonough et al., 1982). Although this device is advertised as excelling at digital filtering and DMA execution, it does neither in Codas. Its primary function is to control the real time signal display of Codas. This processor controls scale, offset, scroll and configuration of up to 16 channels of data in real time. Simultaneous control of the DMA is accomplished by the PC DMA controller. DMA transfers are made in 16 bit words handled by a circular buffer that circumvents the 64 kbyte DMA restriction imposed by the PC. This buffer passes 32 kbytes to disc at a time while accepting the next 32 kbyte section. The maximum size of a data file is 32 Mbytes; this is imposed by DOS (Dataq Instruments, Inc., 1989).

On-board ADC and DAC

The on-board ADC is a self calibrating successive-approximation device with conversion time of 20 microseconds including acquisition time. The ADC input range is +/- 2.5 V.

The on-board DAC has 12 bits of resolution capability with a settling time of 10 microseconds to 0.01 % of full scale transition. The output range is +/- 5.0 V (Dataq Instruments, Inc., 1989).

Personal computer

A Zenith Z-248-12 computer was selected as the host computer for AT-Codas. Equipped with an 80286 16 bit processor and a 80287 math coprocessor, this AT architecture machine runs at a clock rate of 12 Mhz with zero wait states. One Mbyte of RAM, one 40 Mbyte hard disk, dual floppy disks of 1.2 Mbyte (5.25") and 1.44 Mbyte (3.5") capacity handle memory, storage, and backup capabilities. The hard disk is partitioned into 3 separate areas, designated drives C, D, and E. The Codas software, including Advanced Codas, resides on drive D and data are written to drive E before being transferred to floppy disk. This was designed as a safety precaution to minimize the chance of inadvertently deleting or damaging AT-Codas files during acquisition. An analog flat screen color monitor, offering 640 dots X 480 lines of resolution is driven by a Zenith video graphics array (VGA) card. A parallel output port interfaced to a Panasonic KX-P1092i 9 pin printer is available for data reproduction. A hospital grade isolation transformer rated at 4.35 A and 500 VA separates the DAS from room power to eliminate computer generated voltages/currents from creating a ground potential that could endanger the subject during surgery. Also, the transformer offers some protection to the computer against transient voltage potentials.

ANALOG SIGNAL CONDITIONING

Although Advanced Codas can calculate dp/dt and mean pressure, it cannot do so is real time. To display dp/dt and mean pressure in real time requires analog signal conditioning. Also, the pressure signals and possibly the sonomicrometer signal would need to be bandlimited to limit noise and minimize aliasing errors. As the high performance I/O module supplies gain in steps, pressure signals need prescaling to best use the channel input range. A calibrated precision voltage reference was required to accurately scale high-pressure signals. Circuits were constructed to fulfill these requirements.

Circuit Fabrication and Testing

Initial construction by breadboard techniques was used. Circuits for analog signal conditioning were left in the breadboard stage after initial frequency generator and oscilloscope tests were completed. Several experiments were then conducted using the circuits at the front end of the DAS. This allowed for modifications before final construction.

It was decided that a modular approach to construction would allow more flexibility for future studies. One box would contain low-pressure circuitry. One Millar pressure transducer could be input to this box with the resultant outputs of mean and pulsatile pressure signals available. A second box would house the high-pressure circuitry. The signal from a Millar transducer could be input to this box with outputs of two pulsatile pressure signals and dp/dt available. Any of the outputs could be used and further conditioned by either the AT-Codas high performance I/O module or by cascading additional filters. A third box was planned but deemed unnecessary after breadboard testing. This unit was initially designed to be a two-pole low-pass filter to suppress induced noise at the output of the sonomicrometer.

Both circuits were fabricated on Radio Shack universal printed circuit boards. All ground connections were made using 22 gauge insulated wire soldered to a central grounding strip on the board. All jumpers and power connections were made using 24 gauge insulated wire. Dip sockets were soldered to the boards to allow amplifiers to be easily changed. All other components were soldered directly to the board using predrilled perforations.

One percent resistors with 100 ppm/$^{\circ}$ C temperature coefficients were used when possible to achieve the best results from the amplifiers and the best match of resistive components in filters. Panasonic P series two percent polypropylene capacitors and ten percent M series polyester and foil noninductive capacitors were used when possible in an attempt to meet design parameters and avoid drift of the time constant. For larger capacitive values, Panasonic twenty percent tantalum capacitors were used when possible.

Board power buses and the ground strip were connected by wire soldered to female banana binding posts to allow for easy attachment to a separate external regulated +/- 15 VDC power supply. Decoupling capacitors (0.1 uF) were soldered between each bus and the ground strip to eliminate possible power supply noise. Input and output signal connections are made by 1/4 inch phono plugs attached to the side wall of each box.

All signal cables were fabricated using double shielded 4 conductor 22 gauge cable with a stereo-phono plug soldered to one end and solderless banana plugs used at the other end.

After final construction each filter was again tested using a frequency generator and 50 MHz oscilloscope.

Low-Pressure Circuitry

Initial gain

In the first stage an Analog Devices AD524CD precision instrumentation amplifier supplies gain to the differential signal from the Millar transducer system. With the Millar transducer system output of 0.04V/20 mmHg, gain is necessary to split and channel the signal to simultaneously view mean and pulsatile signals. Also, signal gain is necessary to utilize the complete input range of the AT-Codas channel.

Initial DAC low-pressure calibrations utilized the 20 mmHg electronic output from the Millar control unit. However, later uses might require a larger range of measurement and thus a larger input calibration signal to maintain accuracy. The calibration signal should use as much of the channel input range as possible to minimize ADC error. A 15 turn cermet potentiometer was placed in the AD524CD gain loop to allow flexibility of input range and scaling with respect to Codas channel input range. Gain characteristics were tested using a 0.04 V signal from DC to 100 Hz. Gain characteristics remained flat throughout bandwidth tested. Nominal values were as follows:

	Calculated Gain	Measured Gain
Min	61.4X	57.5X
Max	247.9X	230X

Actual gain would allow a prescaled 20 mmHg equivalent calibration output of 230 X 0.04 V, or 9.2 V, to pass to the mean pressure filter and pulsatile output. However, the gain of mean pressure filter would cause saturation if this gain level was used. Using a 20 mmHg equivalent calibration signal at a gain setting of 225X would supply an output of 9 V and would not saturate the low-pass filter. This would be 90% of full scale channel input which would be adequate for calibration. At a low gain setting, a 78 mmHg electronic equivalent of 0.16 V could be input and utilize 92% of the Codas 10 V input range

The AD524CD was selected for its high accuracy/linearity and low drift, offset, and noise characteristics. See figure 1.

Mean pressure filter

Mean low-pressure values are obtained by passing the signal output from the AD524CD through an extremely bandlimited 2nd order Butterworth low-pass filter. A time constant of 2.3 seconds, which corresponds to a corner frequency of 0.07 Hz, was selected to supply a smooth, slowly changing output for real time determinations. The second order Butterworth transfer function used is as follows.

In complex variable s:

$$Av_s/Av_o = 1/((s/w_o)^2 + 2k(s/w_o) + 1)$$

Where w_o = 3 dB point (corner frequency)

k = damping constant

Av_o = stable midband gain

Av_s = input signal

Practical realization of the above transfer function takes the form of the following.

$w_o = 1/RC$

$Av_o = 3-2k$

For a second order system k = 0.707 (Millman and Halkias, 1972). Initial testing of frequency amplitude response provided the following results.

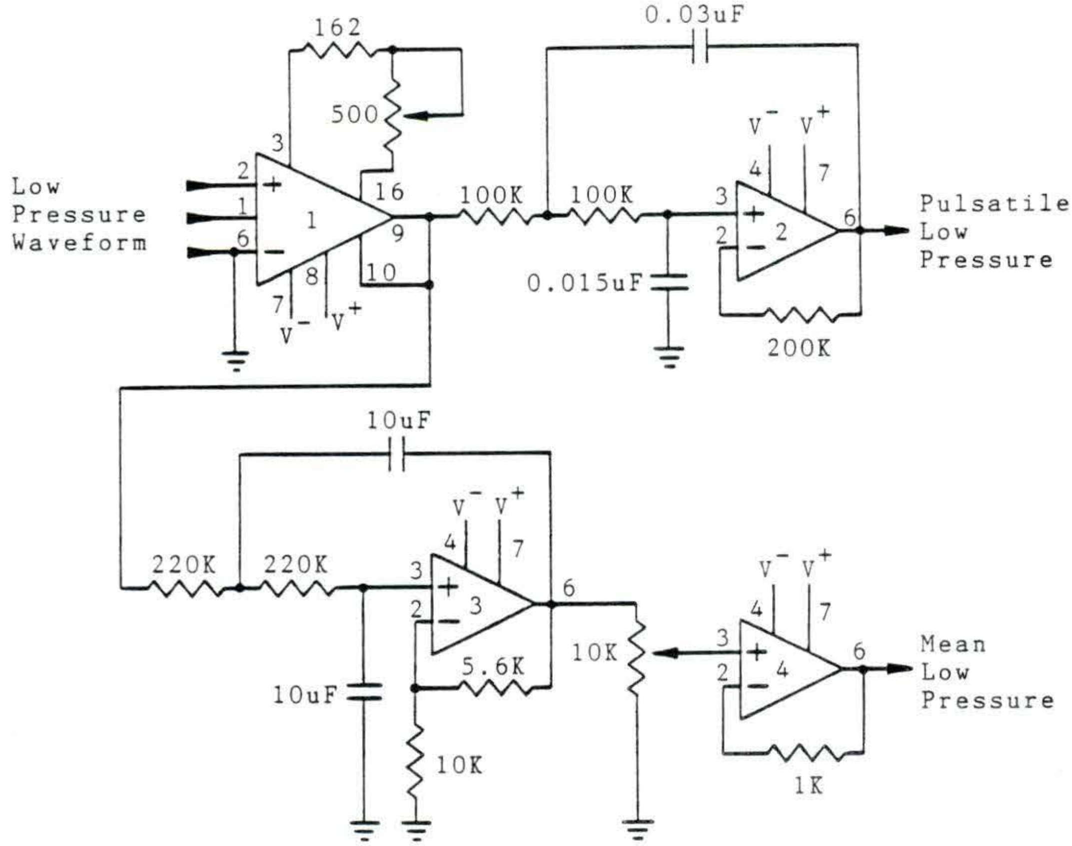
	Calculated Value	Measured Value
3 dB point	0.07 Hz	0.07 Hz
Av_o	1.59X	1.58X

The filter was found to saturate at +14.6 V which limits the input to less than 9.2 V.

The output of the Butterworth filter passes through a follower with adjustable gain. This last stage allows mean low-pressure gain to be set to 1X in the passband and thus scaled during calibration to the pulsatile low-pressure output. Analog devices AD548JN and AD711JN bifet operational amplifiers were selected for use in the mean pressure circuitry because they exhibit excellent low drift/offset characteristics, overall precision, and DC accuracy.

Pulsatile signal conditioning

The second output from the AD524CD is passed through a second order Butterworth low-pass filter to attenuate unnecessary high-frequency components to reduce noise and aliasing effects. An Analog Devices AD544LH bifet amplifier was used because of its excellent low input offset, low drift, low noise and high slew rate characteristics. The filter was designed to have a corner frequency of 75 Hz which corresponds to a time constant of 2 milliseconds. This value was selected as predicted values indicated a flat amplitude to frequency response from DC to within three percent of 37.5 Hz (Coughlin and Driscoll, 1987). This would allow passage of the tenth harmonic of a blood pressure signal at a heart rate of 225 BPM. Midband gain was designed to be 1X. Predicted attenuation of frequencies higher than 75 Hz would occur at -40dB/decade. Initial frequency/amplitude tests were conducted with the input voltage set at 6.6 V to replicate expected values to be input to the filter when measuring pressures in the 0 to 20 mmHg range. In reality, the corner frequency was found between 70 and 71 Hz with a flat amplitude to frequency response from DC to within five percent of 40 Hz. The phase angle was between -100.8° and -102.2° at the corner frequency. The output of frequency components from DC to 100 Hz followed the input signal by 4 milliseconds.



IC Labels Used in Figure 1

IC Label	IC
1	AD524CD
2	AD544LH
3	AD711JN
4	AD548JN

Figure 1. Circuit developed to supply gain, condition pulsatile low-pressure, and determine mean low -pressure

High-Pressure Circuitry

Initial gain

Initial gain is supplied by a differential amplifier built around an Analog Devices AD548JN precision bifet amplifier. Gain level was designed to be 18X to allow for the higher than normal left ventricular pressures anticipated from dogs with experimentally induced cardiac hypertrophy. With the Millar transducer system output at 0.2 V/100 mmHg, a full channel range of 277.8 mmHg could be displayed at this gain setting. This gain would also allow for the signal output to be split and channeled to simultaneously view dp/dt, ventricular pulsatile waveform with a gain of 1X, and the pulsatile pressure waveform with gain of 5X. This gives a better display of the end-diastolic pressure point in real time. See figure 2.

Gain characteristics were tested using an anticipated midrange equivalent value of 0.2 V over a bandrange of DC to 100 Hz. Gain characteristics remained flat throughout producing a gain of 18X.

The AD548JN was selected for its low input offset voltage, low drift, high common mode rejection ratio (CMRR), and excellent DC and AC performance.

dp/dt circuitry

The pressure output from the initial gain stage passes through three additional stages to derive the real time display of dp/dt. The second stage is a second order Butterworth low-pass filter. The third stage is a differentiator developed by trial using Analog Devices application sheets. The fourth stage is an inverting amplifier with adjustable gain. This stage corrects the polarity of output from the differentiator and supplies amplitude adjustment.

The initial differentiator circuit was unstable and did not perform well when breadboarded following the applications specifications. The circuit was sensitive to noise and produced an unstable output with a low signal to noise ratio. This problem was resolved in part by bandlimiting the input signal. In addition, the differentiator resistive and capacitive (RC) values were increased to improve stability. Capacitance was increased by a factor of 10X. Resistor values were then increased above design specifications until output became stable when a 3.6 V peak to peak sine wave was input. Further adjustments of RC values were made to set the timing of the output signal with respect to the input signal. This was done by comparing the output with an input triangular wave signal of known slope. Two switching diodes, biased opposite to one another, were placed in the output loop of the differentiator amplifier to eliminate noise that partially obscured the signal output baseline. As this device allowed for real time impressions, determination of end-systolic and end-diastolic points and not of absolute signal amplitude, it was deemed unnecessary to build a calibration circuit for DAS scaling.

Low-pass filter A second order Butterworth low-pass filter was constructed to bandlimit the input signal. The corner frequency was set by trial to produce a dp/dt signal output that would be usable in real time. Tests showed the corner frequency to be 33.5 Hz with a stable midband gain of 1X. An Analog Devices AD707JN was selected for use in this stage because of its excellent low noise, low voltage offset/drift, and slew rate characteristics.

Differentiator Tests were conducted on the final design to check timing and amplitude of signal output with respect to input signal. Initial amplitude tests were conducted using an electronic signal generator. Using a triangular wave input to the initial gain amplifier and measuring the output of the differentiator before the inverting amplifier with gain produced the following results.

Input Signal	Equivalent Pressure	Frequency	Output
0.2 Vpp	100 mmHg	1 Hz	0.8 Vpp
0.4 Vpp	200 mmHg	1 Hz	1.6 Vpp

0.6 Vpp	300 mmHg	1 Hz	2.4 Vpp
0.2 Vpp	100 mmHg	2 Hz	1.6 Vpp
0.4 Vpp	200 mmHg	2 Hz	3.2 Vpp
0.6 Vpp	300 mmHg	2 Hz	4.8 Vpp
0.2 Vpp	100 mmHg	3 Hz	2.4 Vpp
0.4 Vpp	200 mmHg	3 Hz	4.8 Vpp
0.6 Vpp	300 mmHg	3 Hz	7.2 Vpp
0.2 Vpp	100 mmHg	4 Hz	3.6 Vpp
0.4 Vpp	200 mmHg	4 Hz	6.2 Vpp
0.6 Vpp	300 mmHg	4 Hz	9.5 Vpp

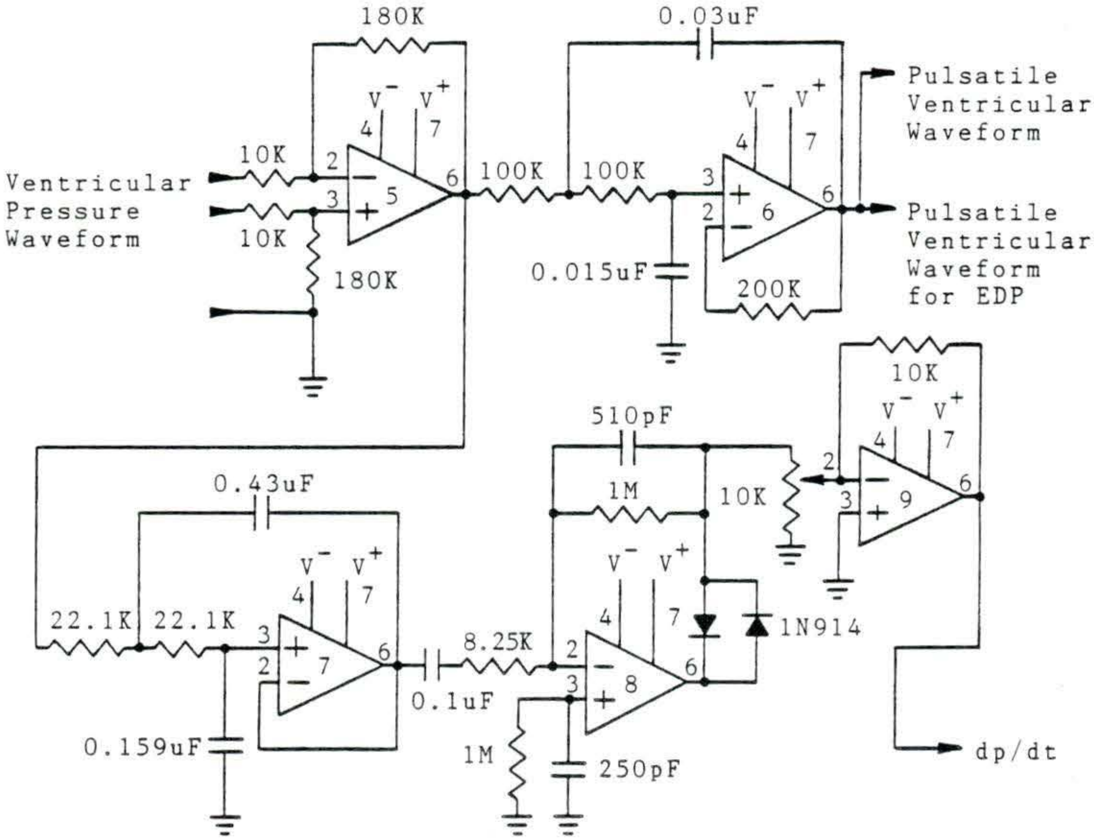
Linearity with scaling of the output, through 3 Hz, would allow for derivative determinations through 160 BPM. By simple division it can be seen that the output does replicate the derivative process. For example at 1 Hz an input of 0.2 Vpp would rise from 0 to the positive peak in 0.25 seconds. This would be a positive change from 0 of 0.1 V in 0.25 sec or 0.4 V/sec. Using Millar system specifications, the pressure equivalent is 200 mmHg/sec.

Additional tests were conducted with a Millar multifunction pressure generator coupled to a Millar transducer system input to the initial gain amplifier. Output from the inverting amplifier with gain was compared to the input signal on a storage oscilloscope. Values were tested from 1 to 5 Hz replicating 60 to 300 BPM. The Sampling rate was set at 1000 Hz/channel on the oscilloscope. At this sampling rate it was found that there is no delay in the start of the rise of the derivative signal compared to the start of rise of the input square wave. There was no delay in the start of drop of the derivative signal compared to the start of drop of the input square wave. This makes this signal very usable for determination of end-systolic and end-diastolic points in real time. There was a delay of 20 milliseconds between reaching the peak derivative and reaching the peak level of input signal. This is inconsequential in the GIK study.

Inverting amplifier with gain Output from the differentiator is inverted with respect to the input. This stage corrects polarity of output so that a positive rise in the pressure signal will produce a positive spike at the dp/dt output. A 10 kohm potentiometer at the input of the inverting circuit, built around an Analog Devices AD707CH, supplies adjustable gain to the output. Gain can be set to best optimize display of dp/dt.

Pulsatile signal conditioning

The second output from the AD548JN is passed through a second order Butterworth low-pass filter to attenuate high-frequency components in order to reduce noise and aliasing effects. Output from this filter is split with one signal used to display pulsatile left ventricular pressure and the second output going to the AT-Codas high performance I/O module where it received a gain of 5X for the real time display of the EDP point. The filter was designed to have a corner frequency of 75 Hz with a stable midband gain of 1X. Initial tests showed the corner frequency to be 77 Hz and stable midband gain to be 1X. Amplitude to frequency response remained flat from DC to 35 Hz and flat to within 5% at 45 Hz. This allows adequate passage of the 10th harmonic of a pressure waveform at a heart rate under 270 BPM or of the 20th harmonic at a heart rate under 135 BPM. Rolloff rate was found to be close to the design specification of 40 dB/decade. Phase angle was approximately 110° at the corner frequency. Output of frequency components from DC to 95 Hz followed the input signal by 4 milliseconds. An Analog Devices OP27GH was selected for use in this circuit because of its low offset voltage stability, high stability over time, and high slew rate characteristics.



IC Labels Used in Figure 2

IC Label	IC
5	AD548JN
6	OP27GH
7	AD707JN
8	AD548KN
9	AD707CH

Figure 2. Circuit developed to supply gain, condition two pulsatile high-pressures, and determine dp/dt

SYSTEM PERFORMANCE IN THE GIK STUDY

Overview

The GIK study was comparitive in nature. A thoracotomy was used to expose the heart in anesthetized dogs. The Millar transducers were placed by stab wounds. The high-pressure Millar #1 was placed in the left ventricle and the low-pressure Millar #2 was placed in the left atrium; both were sutured into place. The sonomicrometer crystals were placed in the free wall of the left ventricle proximal to the apex. The crystals were positioned facing one another approximately 10 millimeters apart. Subcutaneous EKG electrodes were attached to the dog's limbs and connected to the Datascope 861 monitor and linked to the DAS by patch leads. Hemodynamic and myocardial mechanical data were recorded after the heart and transducers had been allowed to stabilize. Cooling and heart-lung bypass would then ensue for one hour. After cessation of bypass and beginning of normal heart activity, data were again recorded. Data were sequentially recorded at 15, 30, 45, 60, and 90 minutes from the time of weaning from the heart-lung bypass pump. Data were displayed throughout periods before and after bypass to assess heart performance and to attempt to control preload by infusion or removal of blood volume. After the last recording was made, required digital signal processing was completed using the Advanced Codas software and the data were reviewed. Percent recovery was then calculated by using the dog's pre-bypass cardiac data as the control values for comparison to post-bypass data.

Calibration preparation

Equipment was unplugged from room power until a majority of the electrosurgery used for the thoracotomy was completed. This precaution was taken to avoid possible transient damage to the DAS equipment. The computer, sensor devices, and analog signal conditioning circuits were then powered up for approximately thirty minutes prior to calibration to allow for temperature stabilization of all electronic components. The Millar transducers were protected from light and allowed to soak in sterile deionized water for one half hour for stabilization before calibration. A room temperature limit of 75° F was set to avoid heating problems with the DAS equipment. If, at any time defibrillation was required, DAS equipment was turned off or to standby, unplugged from room power, and if possible disconnected from the animal to avoid possible equipment damage from the large DC pulse.

Data file

Prior to calibration a control data file was created on hard disk drive E. File size for the GIK study was normally specified to be 720 kbytes for reasons explained below.

Usually 3 floppy-disk backup copies of each file were created. For convenience of review, each dog's data files were left on drive E for two days after recording. Files would then be deleted from the hard disk to make room for new files.

Channel configuration

Creating a data file simultaneously opens AT-Codas to the setup operating mode. As one of four operating modes, setup is used to configure data acquisition parameters such as channel configuration, throughput rate, and I/O amplifier gain/offset. Other modes of operation are standby, record and post acquisition. Once the record mode has been initiated for a file, the setup mode cannot be re-entered and data acquisition parameters are fixed until the file is closed. Once configured data acquisition parameters have been established, they may be set to the default condition for future use. This is done by the AT-Codas enable default condition command.

For the GIK study, eight Codas channels were activated and configured for operation using the AT-Codas configure channels command. With signal conditioning, it was possible to set all channels for single ended operation. The AT-Codas screen annotation function was used to specify channels for ease of identification during the real time display. The following configuration was used for display and recording of parameters on the DAS. See figure 3.

Channel	Parameter	Device
1	Pulsatile LVP	Millar system #1
2	Pulsatile LAP	Millar system #2
3	LVEDP	Millar system #1
4	Analog dp/dt	Millar system #1
5	SS	Sonomicrometer
6	Subcutaneous EKG	Datascope 861
7	Analog MLAP	Millar system #2
8	Endocardial EKG	Sonomicrometer

The AT-Codas display formats allow for channels to be viewed in a variety of ways. Display format #7 allows all eight channels to be displayed at once with no overlap. For a better resolution of the display, format #1 allows for any one channel to be displayed. This was useful in real time display when precise pressure determinations were required.

Throughput selection

Initial throughput requirement estimates were based on the EKG. However, a change in protocol made faithful reproduction of the EKG unnecessary. Both EKG signals were used for display and general assessment of cardiac electrical

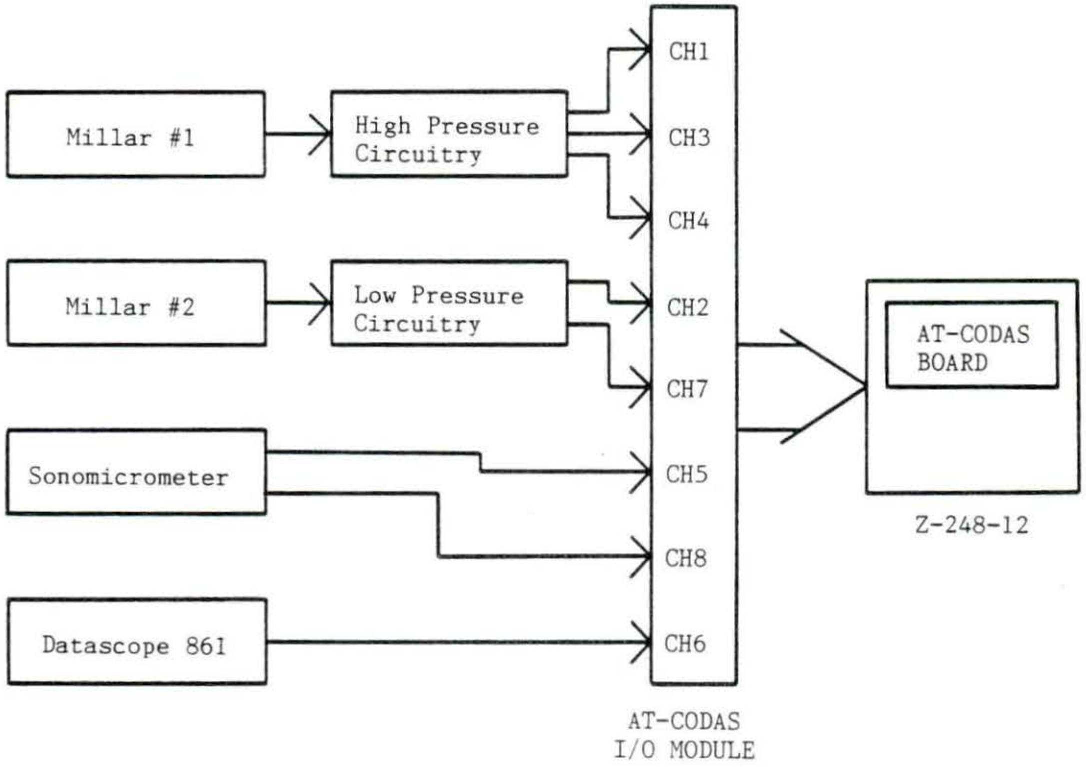


Figure 3. Block diagram of the DAS as it was configured for the GIK study

activity in a clinical fashion. Left ventricular pressure for determination of dp/dt thus became the most demanding signal with respect to the throughput rate. Estimating the bandwidth requirements of the signal to be between 30 and 50 Hz would set the minimum throughput at 100 Hz/channel. Higher rates would be desirable to increase resolution and avoid aliasing effects. A compromise of throughput speed, storage space, and operational function had to be made. Trials were conducted to determine the best combination.

Rate limits were imposed by file size. To insure that data were not lost, it was decided that floppy disk backup files would be made by using the DOS copy command immediately after recording each data set. It was determined that the DOS backup command would eliminate a large time interval of real time display and would be undesirable for use. Thus each file would need to fit on a single 1.2 Mbyte floppy disk. The best combination tried set throughput to 2000 Hz and the data file size to 720 kbytes. This allowed for over three minutes of data collection with each individual channel having a throughput of 250 Hz. At this rate all data collected plus all required calculated data channels could be stored on a single 1.2 Mbyte floppy disk. For cases that required more data collection a file of 1170 kbytes could be created at the same throughput and stored by the DOS copy command on a single 1.2 Mbyte floppy disk. This would allow for approximately 5 minutes of data collection; however, it would leave no space on the floppy disk for calculated data channels. Hard disk space was required to calculate and display additional channels. The AT-Codas waveform compression function was used to slow down the real time display while not affecting storage of data to disk.

When designating data file size, it should be noted that AT-Codas requires 2 bytes of storage space for each sample collected. Eleven hundred and fifty-six bytes of each data file is always used to write a file header which contains 29 elements of recording constants. In addition, AT-Codas reserves two separate 64 kbyte field locations for event markers and event comments. One marker without comment is automatically written every time data storage to disk is started. Additional markers may be written with or without comment. Markers without comments require 4 bytes of space, markers with comments require 8 bytes of space plus 1 byte for each character used to write the comment, and event markers with comment and date and time stamping use 16 bytes of space plus one byte of space for each character used while writing the comment. It is possible that extensive use of event markers and/or comments could increase file size beyond floppy disk storage capability.

Calibration

While still operating in the setup mode, AT-Codas display format one was used to visually set the electronic zero of both Millar pressure transducers. Transducer zero was adjusted by the potentiometer on the Millar TCB-500 control box. Channel baseline offsets were adjusted to desired levels by the amplifier offset function and the gain of the EDP channel was set to 5X by the amplifier gain control function. Gain at the last stage of the low-pressure mean average filter was checked and adjusted, if necessary, in order to match the output of the low-pressure pulsatile signal. Sonomicrometer offset and gain levels were checked and adjusted.

AT-Codas acquire to disk mode was then selected and the precision voltage reference electronic calibration standards were entered as efficiently as possible to avoid wasting data file space. EKG signals and analog dp/dt signals were left uncalibrated. The event was marked with comment for later identification and the standby mode of operation was then selected.

The shell to post command function was activated to temporarily enter the post acquisition mode and enter engineering units. The cursor oriented calibration function allowed user selected engineering units to be affixed to both high and low calibration standards of all selected channels by simple keyboard commands. Displayed maximum, mid, and minimum values were then recorded on paper to compare with previous and future values to check for possible error and repeatability.

Recording pre-bypass data

After calibration, shell back to real time Codas would bring the system to the standby mode of operation with real time display scaled to the calibration units selected. Maximum, mid, and minimum channel values were still available for display in real time if so desired. The system was left in the standby mode for display until it was time to record control parameters. Once the control data was recorded, with event marker and comment, shell to post would be invoked to quickly check the quality of the data. Shell to DOS would then be invoked to make two floppy disk backup copies of the recorded data.

Digital conditioning

All Advanced Codas conditioning calculations have to be accomplished after closing the data file. Attempts to calculate values before doing so resulted in errors that completely destroyed the data collected. The exit Codas command will return the user to DOS, close the data file, and allow for safe use of the Advanced Codas features. For the

GIK study, 3 channels of data were digitally conditioned by the Advanced Codas techniques.

EKG conditioning

The EKG display from the Datascope 861 leads was smoothed by using the Advanced Codas waveform moving average feature. The minimum smoothing factor of three was usually sufficient to eliminate induced noise and produce a smooth waveform. The smoothed EKG signal was then displayed on channel 11. Higher smoothing factors were used to remove noise if necessary; however, too much smoothing resulted in waveform distortion.

Digital MLAP

By selecting a higher smoothing factor, the waveform moving average feature was used to produce a good first order approximation of mean left atrial pressure. The Advanced Codas formula used to calculate the mean average is shown below.

$F_m = (1.6 \times CSN)/N$

Where CSN = channel sample rate

N = number of smoothing factor

F_m = bandwidth of mean signal in Hz

For a sampling rate of 250 Hz, a smoothing factor of 400 produced a good mean signal. By always using the same smoothing factor, the digital mean was initially considered more repeatable than its analog counterpart. It was thus used to make any MLAP signal amplitude determinations during the GIK study. In reality, the MLAP produced by the analog and digitial conditioning usually varied less than +/- 0.4 mmHg from one another. The digital technique did show an advantage over its analog counterpart in times of rapid change, when the long time constant of the analog mean-average filter would produce a slight time delay. Advanced Codas would automatically scale this signal to the channel from which it was calculated and then display the MLAP with the appropriate units on channel 10.

Digital dp/dt

dp/dt was calculated for amplitude and timing determinations by use of the derivative feature. This feature has a built in smoothing factor that may be selected to attenuate noise present on the signal to be calculated. The smoothing factor selected will determine the number of points evaluated for the derivative slope. A smoothing factor of six produced a very usable derivative output. Digital dp/dt was scaled to the channel from which it was calculated and it was displayed on channel 9.

Bypass display

A file was created and AT-Codas activated for display while the dog was on heart-lung bypass. Channels were not calibrated. The AT-Codas display was used to determine if the heart attempted to resume activity before cessation of bypass.

Recording post-bypass data

DAS components were powered up and all sensors connected to the animal after routine post-bypass defibrillation. Channel configuration and calibrations were completed as before bypass. Display maximum, mid, and minimum channel unit values were recorded on paper and compared to pre-bypass values as a check for error and repeatability. Data were then sequentially recorded for a maximum of 90 minutes from time of weaning from the bypass pump. Two floppy disk backup copies were made and updated after each event. After closing the data file, Advanced Codas routines were used to condition the data as before. After digital conditioning, a third floppy disk backup copy of each file including calculated data was made.

Calibration check

The Millar pressure transducers were cleaned, covered from light, and soaked in sterile distilled water for one half hour after surgery. The drift from zero of each transducer was checked and recorded. Over the interval checked, overall maximum drift of the low-pressure Millar #2 with amplification was found to be less than +/- 0.5 mmHg. Overall drift of high-pressure the Millar #1 with amplification was found to be less than +/- 1 mmHg. Voltage references used as pressure equivalents for standards were periodically checked and calibrated against Millar units coupled to a water column with a maximum of 170 mmHg.

<u>Data review</u>

Data were reviewed on the same day as the experiment for evaluation. Values of interest were recorded on paper and data examples printed using the Codas hardcopy utility.

Results

The following is a set of data examples produced by the AT-Codas hardcopy utility. Figure 4 shows 7 channels of data taken from a dog prior to heart-lung bypass. Throughput rate is 250 Hz/channel limiting the time divisions to 4 milliseconds. This supplied the needed resolution to discriminate between data points of interest in this study. Compression factor of 2X was used to produce the hardcopy, limiting the time divisions to 8 milliseconds. The heart rate at the time of the recording was 98 BPM which is much less than the study average maximum of 125 BPM. Channels displayed are as follows from top to bottom. All parameters highlighted with an asterisk were calculated data channels.

Channel	Parameter	Device	+/- Full scale	Mid scale
#1	LVP	Millar #1	+/- 69.2 mmHg	48.1 mmHg
#2	LAP	Millar #2	+/- 7.54 mmHg	8.26 mmHg
#3	LVEDP	Millar #1	+/- 27.6 mmHg	16.6 mmHg
#9	dp/dt*	Millar #1	+/- 2160 mmHg/s	324 mmHg/s
#5	SS	Sonomicrometer	+/- 2.13 mm	7.02 mm
#8	EKG	Sonomicrometer	+/- 1.25 V	0.375 V
#10	MLAP*	Millar #2	+/- 15.1 mmHg	10.5 mmHg

Figure 5 shows two channels of the same pre-bypass data displayed in figure 4. In this example sonomicrometer and calculated dp/dt data were printed alone by AT-Codas to supply better resolution. Scaling and configuration for this example are as follows.

Channel	Parameter	Device	+/- Full scale	Mid scale
#9	dp/dt*	Millar #1	+/- 2160 mmHg/s	324 mmHg/s
#5	SS	Sonomicrometer	+/- 2.13 mm	7.02 mm

Figure 6 shows calculated dp/dt and sonomicrometer data from the same dog at 45 minutes after weaning from bypass pump. Recording rate and channel compression display are the same as in figure 4. Heart rate was 112 BPM at the time of recording. Perpendicular lines indicate end-diasolic (A) and end-systolic (B) points, as determined by dp/dt, to determine percent SS of the free wall of the ventricle. Scaling and configuration for this example are as follows.

Channel	Parameter	Device	+/- Full scale	Mid scale
#9	dp/dt*	Millar #1	+/- 2160 mmHg/s	0 mmHg/s
#5	SS	Sonomicrometer	+/- 2.13 mm	702 mm

Values collected from these two events produced the following information.

Parameter	Pre-bypass	Post-bypass 45 min
EDP	10.7 mmHg	9.1 mmHg
LAP	6.8 mmHg	6.9 mmHg
EDL	8.7 mm	7.3 mm
ESL	5.4 mm	4.8 mm
SS	38 %	34 %
dp/dt(positive max)	2268 mmHg/s	1779 mmHg/s
LVP (peak value)	100 mmHg	84 mmHg

In this example percent recovery was based in part on a comparison of before and after bypass values of SS and dp/dt.

In this example the SS percent recovery was 90% while the raw dp/dt recovery was 52%. The usefulness of the GIK solution in improving myocardial preservation during heart-lung bypass has not yet been determined.

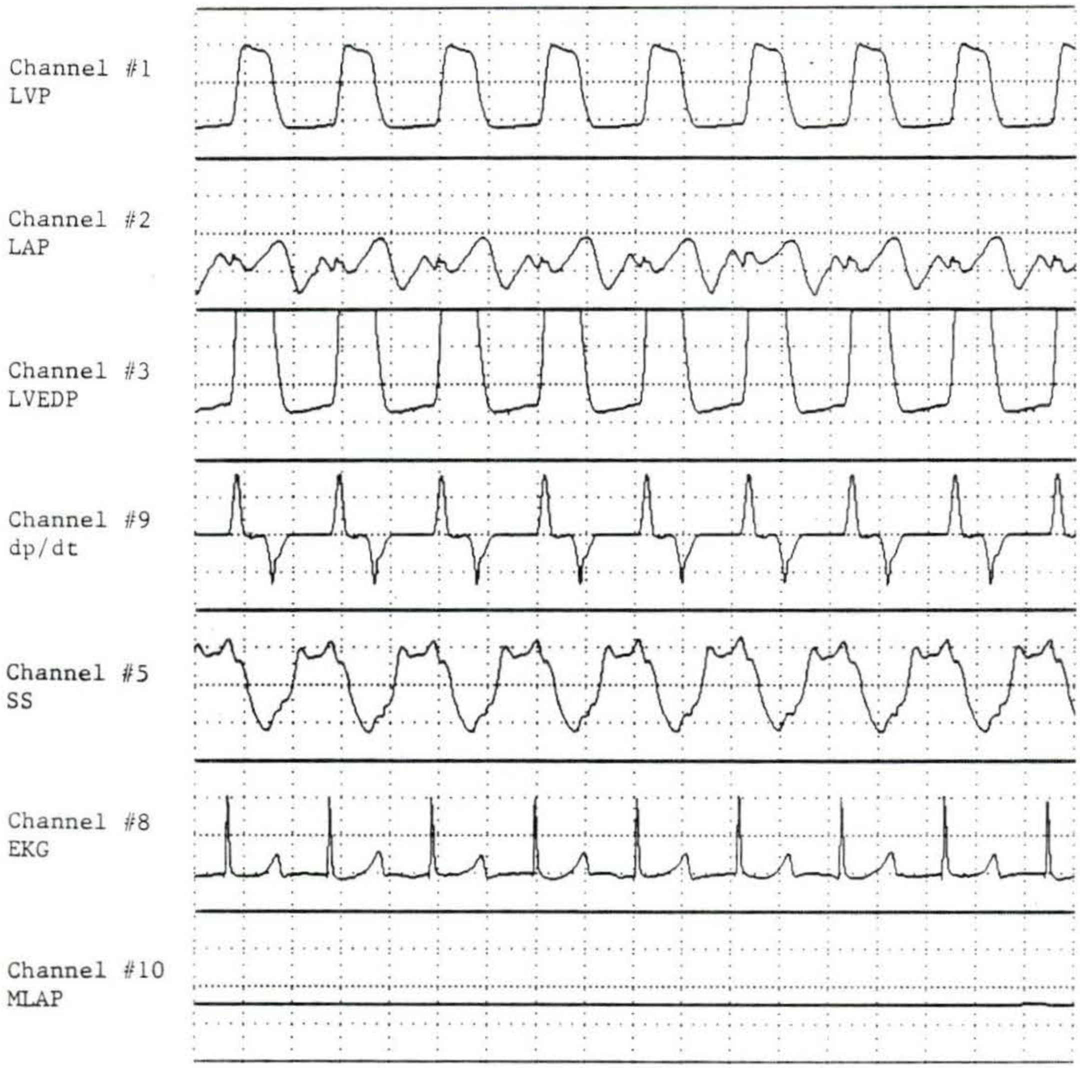


Figure 4. Seven channels of data taken from a dog prior to heart-lung bypass

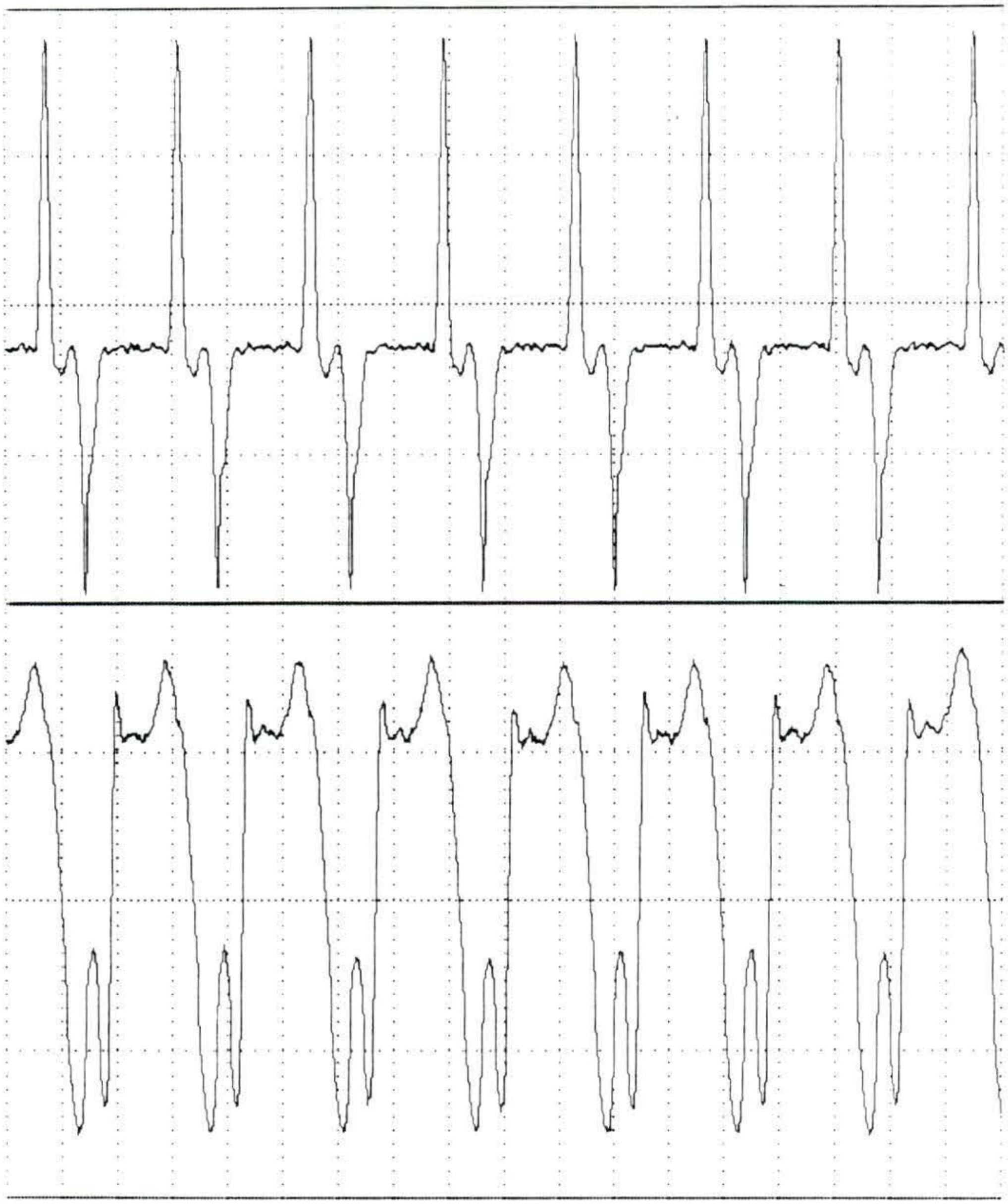


Figure 5. Two channels of the same data displayed in figure 4: calculated dp/dt and sonomicrometer

Figure 6. Calculated dp/dt and sonomicrometer data taken from the same dog at 45 minutes after weaning from bypass pump

RECOMMENDATIONS FOR FUTURE DEVELOPMENT

In light of my experiences with AT-Codas and the Iowa Methodist Health Foundation studies, I would do the following to improve the quality of data acquisition for future work of the same nature.

I would provide a new signal conditioning unit for each input channel of AT-Codas I/O module. I would have the device adjustable in order to easily vary passband on a channel basis without phase shift between channels. I would mount the device as close as possible to the AT-Codas I/O module to insure induced noise would be kept to a minimum. I would supply a rolloff rate of at least 80 dB/decade to further decrease the effects of aliasing.

I would install signal analysis software such as Dadisp to the same hard drive that holds the AT-Codas software. I would remove software other than utilities that would be used for data acquisition/conditioning, data analysis, report/figure generation, and communication from the hard disk. This would open more space for acquiring and manipulating data and decrease chances of software conflicts and errors. I would have communications connection available in the vicinity of the data acquisition to allow for possible sharing/shipping of data and communication with libraries and others conducting similar research.

I would conduct signal frequency analysis before the start of a study to confirm bandwidth requirements and to spot and eliminate possible noise sources.

I would provide an adjustable precision voltage reference that could supply and display output to all AT-Codas channels simultaneously to speed up calibration and avoid storage waste. Also, I would provide a portable 250 mmHg mercury column for confirmation of voltage equivalents.

Finally, I would explore the capabilities and use of compatible analysis and report generating software such as Asystant, Asyst, Dadisp, and Lotus to enhance the capabilities of the AT-Codas DAS.

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