MEASUREMENT OF THE PRE-EJECTION PERIOD FROM THE ESOPHAGUS

by

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ABSTRACT

In this thesis, an algorithm is developed to measure one of the systolic time intervals, the pre-ejection period, from ECG and electrical impedance signals available in the esophagus. The signals are processed by a microprocessor system. These signals are produced by an esophageal probe monitoring system being developed by the Bioengineering Unit of the Harvard Anesthesia Center at the Massachusetts General Hospital. A correlation detection routine is used to find the fiducial point on the impedance signal that indicates the end of the pre-ejection period, and was found to be very effective. With this correlation routine it is a simple matter to change the fiducial point detected by changing the template in the system's memory.

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Chapter 1

INTRODUCTION

1.1 Aim of this Thesis

The Bioengineering Unit of the Harvard Anesthesia Center at the Massachusetts General Hospital is currently developing a patient monitoring system for use during anesthesia and recovery based on an esophageal probe for unified signal acquisition, and a microprocessor based console for signal analysis, display, and recording. It appears that one parameter characteristic of cardiac function, the pre-ejection period (PEP), is measurable from esophageal ECG and impedance signals. The purpose of this thesis effort was to develop an algorithm to measure the pre-ejection period by processing the ECG and impedance signals with a microcomputer system.

1.2 Systolic Time Intervals

A need exists for a relatively safe and non-invasive measure of myocardial function of the anesthetized or critically ill patient. Systolic time intervals, though not providing a true index of myocardial contractility, have been shown to provide useful information about trends in cardiac performance.

A number of systolic time intervals have been defined:

1) Pre-ejection period (PEP) - onset of ventricular depolarization (Q-Wave) to the opening of the aortic valve.

2) Left ventricular ejection time (LVET) - opening of
the aortic valve to its closing.

3) Isovolumic contraction time (ICT) - onset of left ventricular pressure rise to opening of the aortic valve.

4) Total electro-mechanical time (EMT) - onset of ventricular depolarization (Q-Wave) to aortic valve closing. Aortic valve closing is commonly detected by the second heart sound (S2).

Both PEP and LVET have been used as a measure of myocardial function after myocardial infarction with varying amounts of success (1). Rasmussion and Sorenson (2) used the PEP/LVET ratio as an index of cardiac function during halothane and fluoroxyene anesthesia. Also Smith et al (3) have found the inverse of the square of the PEP to correlate well with peak ascending aortic blood flow acceleration. Metzger, et al. (4) have described the relationship between the true isovolumic contraction time and the PEP and found that the latter adequately reflects changes in the ICT, which is probably the most direct time-interval index of contractility.

The most widely used technique for measuring PEP and LVET currently, uses the ECG, phonocardiogram, and arterial pressure. EMT is derived from the ECG and the phonocardiogram, and the LVET is obtained from the arterial pressure curve as shown in fig. 1.1. PEP is then found by the relation PEP = EMT-LVET. Though this technique is relatively non-invasive it does pose some problems. If the arterial pressure is measured in a peripheral
Fig. 1.1 - Systolic Time Intervals
vessel the risk of serious injury is small, but the actual waveform seen may be seriously distorted from that of the aorta. If on the other hand, a more central artery such as the carotid is used, the pulse wave shape is much less distorted, yet the risk of injury is higher. Perhaps the most serious drawback is that this measurement requires up to four separate instrument packages. One instrument for each parameter, and one instrument for computing the systolic time intervals. Additionally, the measurement requires at least three sets of wires going to three transducers on the patient.

1.3 The Esophageal Probe

Currently used monitoring equipment is, in general, poorly designed from a human engineering standpoint. Each instrument is in its own package, and has its own power supply, sensors, etc. Even with multiple functions in one package such as in a multichannel physiological recorder, each additional measurement requires more wires to the patient, and more controls (probably with differing standards) adding to an already cluttered environment. This situation motivated the effort described above to develop a monitoring system based on an esophageal probe for gathering information and a microprocessor based console for gathering, analyzing, and displaying the data. It is hoped that this monitoring system will solve the above problems by taking a systems approach to the problem of monitoring. In addition it is
hoped that more information can be derived from a coherent presentation of the measurements than from all of them separately. That is, the value of the whole should be greater than that of the sum of its parts.

The esophageal stethoscope is widely used today as a monitoring device during surgery. It is simply a flexible plastic tube with a balloon on the tip that acts as a transducer sending heart and breath sounds up the tube directly to the listener's ear. The esophageal stethoscope remains in place well over the course of an operation and is out of the sterile field. Because of the proximity of the esophagus to the heart and lungs, the addition of a few sensors could produce a wealth of information about the cardio-pulmonary system. Sensors that could be added include a thermistor for core temperature, impedance electrodes to measure fluid shifts, a microphone for heart and breath sounds, ECG electrodes, doppler crystals for blood flow measurements, and a transducer for intrapleural pressure measurements. In order to choose which sensors to add, one must consider how useful the information is, how well the measurements complement each other, and how expensive the measurement is.

1.4 Impedance Measurements

One of the easiest sensors to add to the probe is a set of impedance measuring electrodes. By passing a small alternating
current (20-100 kHz.) through the body via two electrodes and measuring the voltage produced, either via the same electrodes or two other electrodes, one may measure the impedance of the body tissue. Changes in this signal are caused by shifts of fluid, of a different resistivity from that of tissue, into or out of the region being measured. For example, the regional impedance changes as blood moves into and out of the major vessels and the heart, and as air moves into and out of the lungs. Researchers since the 1930's have met with only limited success in measuring absolute volume shifts in fluids. In order to measure such things as tidal volume, cardiac output, pulmonary edema, and total body water, it seems that relative volume information and timing information about local fluid shifts may be much more reliable. Measuring thoracic impedance can give us this relative volume, and timing information. In addition it has been shown that impedance measurements from the esophagus produce much more information than traditional extrathoracic measurements (5,6).

1.5 Microprocessors

Until recently, small instruments either had to be limited to rather simple analog and/or digital computing power, or be connected in some way to a central computer by telephone line or other means. With the advent of microprocessors (a complete small-scale computer in the form of a few LSI integrated circuits) an instrument may now be given a large amount of
computing power without having to be connected to a large system with all its problems (connection, down time, expense, etc).

This computing power allows great flexibility in the choice of sensors since greater signal-processing capability is available. This expanded capability also allows more human engineered display implementation, intelligent data logging, trend analysis, and much more sophisticated alarm algorithms to be implemented.
Chapter 2
INSTRUMENTATION

2.1 Summary

This chapter includes descriptions of the instrumentation used to record the needed signals from the esophagus of a dog, the probe used, and the microprocessor development system used to develop detection algorithms.

2.2 The Probe

The esophageal probe (fig. 2.1) used in this thesis was constructed by Richard Scordato (5,6) for his master's thesis done in this lab in 1975. He described its construction as follows:

The probe was modified from a commerical esophageal stethoscope that was made from plastic tubing 6 mm in diameter and 54 cm long. Ten electrodes were added to the probe. The electrodes were fabricated by wrapping four turns of silver wire around the probe as shown in (fig. 2.1). Connection was made to each of the silver electrodes by soldering #30 insulated copper wire to them, and threading these lead wires through the tubing. The electrodes were numbered sequentially from one to ten starting at the distal end of the probe. All electrodes were spaced 2 cm apart except for electrodes 2 and 3 which were 1.7 cm apart; 3 and 4 which were 2.5 cm apart; and 4 and 5 which were 1.8 cm apart. This uneven spacing was due to the balloon pick-up for the phonocardiogram which was placed between electrodes 3 and 4.

Scordato chlorided the silver electrodes for his later
Fig. 2.1 - The Esophageal Probe
experiments. Although the silver chloride coating was intact for all his experiments, it was not intact for the recording done for this thesis.

All impedance measurements made used the four-terminal method. This was done to reduce the effect of contact impedance. If the contact impedance of the electrodes is modeled as a pure resistance, and the body tissue is modeled this way also, then the models of a two-terminal versus a four-terminal measurement in fig. 2.2 apply. If the input impedance of the voltage measuring device is high, the effect of contact impedance can be nullified by the four-terminal method.

The electrode numbering system adopted by Scordato is continued in this thesis. That is, four numbers are given consecutively, such as 1234. The two outer numbers (1 and 4) signify the current electrodes, and the two inner numbers (2 and 3) signify the electrodes used for voltage sensing.

Scordato did both experimental and theoretical work on the field produced and the volume sensed by several electrode configurations. The experimental work was done with the probe in a 25 cm diameter plastic container filled to a depth of 20 cm with 0.9% saline solution. He found the field produced to be essentially a dipole field for electrode spacings smaller than the radius of the container. He also found that the diameter of the region sensed by 4 consecutive electrodes (e.g. 1234) was roughly the spacing of the electrodes (2 cm in this example). This relationship held for larger spacings as well.
Fig. 2.2 - Two and Four-terminal Electrode Models
The position of the esophageal probe in a dog can be clearly seen in the X-Ray reproduced in fig. 2.3. The bright object near the diaphragm is the plug at the distal end of the probe. The electrode nearest the bright spot is 1.

Scordato found that the probe could be positioned quite accurately by looking at the ECG as measured from electrodes 4 (positive terminal) and 5 (negative terminal). As electrodes 4 and 5 were moved past the heart dramatic changes in the morphology of the signals were seen. By moving the probe until the ECG from electrodes 4 and 5 had a positive P-wave and a bi-phasic QRS complex, the probe could be reproducibly positioned such that electrode 4 and 5 were directly behind the heart. The probe could then be taped to the jaw of the animal to hold the probe's position for the remainder of the experiment.

2.3 Data recording

For the purpose of this thesis, it was necessary to record on magnetic tape for later reproduction the carotid pressure, ECG and impedance information from the esophagus of a dog. In this section, I will describe the instrumentation used to do that recording.

The current source used for the impedance measurement is a Hewlett Packard (H.P.) 241A signal generator coupled to the current electrodes through an isolation transformer and a 100
Fig. 2.3 - X-Ray of Probe in Dog
kohm resistor. It delivers a current of 50 micro-amps.

The voltage from the inner electrodes was then measured with a Princeton Applied Research (PAR) 128 lock-in amplifier. This instrument has an input impedance of greater than 55 kohm at 100 kHz and a CMRR of greater than 100 dB. Its bandwidth was set at 150 Hz. The PAR 128's output was coupled to the H.P. recording system via an H.P. 8802A medium gain amplifier.

The carotid pressure was measured using an H.P. 8805 carrier amplifier. The ECG was taken using an H.P. ECG amplifier.

All the information then went into an H.P. 7418A direct writing recorder which was coupled to an H.P. 3900 FM tape deck. The data from a day long experiment could then be recorded on magnetic tape and notes made on its audio track with intermittent records made on the strip chart output.

2.4 Software Development System

The Bioengineering Unit at the MGH has assembled a powerful microprocessor development system for the Intel 8080. This system consists of an Intellec MCS 8/MOD 80, a CRT terminal, a 30 cps hard copy terminal, a paper tape reader and punch, a cassette magnetic tape unit, a floppy disc, and a 300 baud modem. Resident software includes a text editor, an assembler, a system monitor, and a floppy disc operating system.

The Bioengineering Unit has also developed a standard bus structure with standard CPU and memory cards for use of the 8080
In medical instruments. A major feature of this bus structure is that one output of the Intellec system is a card that mimics the actions of a CPU card in an instrument. It is thus possible to edit and assemble a program on the Intellec system, and then debug it in the environment of the actual instrument. A picture of the Intellec and the microcomputer hardware of an instrument under development are shown in fig 2.4.

As another major programming aid, the Bioengineering Unit has installed a high-level language (PL/M) cross-compiler for the 8080 on the Multics system. It has been established that the average output of a professional programmer is ten lines of tested and debugged code a day no matter what the language (6). Also, it tends to some extent to be much easier to keep track of the overall scope of a program in a high-level language. A high-level language tends to be self documenting (7). For prototype instrument development, or to test the feasibility of an idea, PL/M or high-level languages in general can be most useful. During the course of this thesis it was discovered that the software implementation of the algorithm was of sufficient magnitude to warrant the use of PL/M.

2.5 Hardware Constructed

Although the primary goal was to develop an algorithm for computing PEP, a certain amount of hardware had to be designed and built to bring the necessary analog signals into the Intellec
Fig. 2.4 - Microcomputer Development System
In digital form, and to display that and other information in a meaningful form. To this end, three cards were built to fit the standard bus described above. They were an eight-channel ten-bit a/d card, a three channel d/a output card, and a card to generate a 500 Hz interrupt signal for sampling timing. Also, an analog preprocessor was fabricated to provide gain and offset, and to low-pass filter the input signal from the H.P. tape deck. This preprocessor unit also provided ground isolation between the Intellec and the H.P. tape deck to reduce noise problems. Figure 2.5 shows the block diagrams of these circuits.

The three eight-bit O/A outputs were used to drive the X, Y, and Z axis inputs of a Fairchild 737A CRT display. By outputting a point at a time, at software limited speed, it was possible to repetitively display approximately 850 dots without flicker. This allowed up to three 256 point graphs to be put on the screen along with a few indicator arrows. The use of this unit for display of input data, intermediate results, and the results of searches for fiducial points was of immense help in algorithm development.
Hardware Constructed for Thesis

Fig. 2.5 - Block Diagram of Hardware Constructed
Chapter 3
ALGORITHM DEVELOPMENT

3.1 Summary

In this chapter the nature of the signals to be processed will be discussed in section 3.2. In section 3.3, I will attempt to specify the problem more clearly and present the solution derived. Section 3.4 gives a description of the program implemented, and section 3.5 describes the results of testing done with that program.

3.2 The Signals

The ECG, carotid pressure, and the impedance were recorded from a 15 kg male mongrel dog on October 30, 1975. The impedance signal that seemed most promising was that obtained from electrode combination 4567. A strip chart output of these signals is shown in figure 3.1 Although a tape recording suitable for use during algorithm development and testing was only made from one dog, many other paper records were obtained by Scordato in his master's thesis work and were comparable to those obtained here on tape.

In fig. 3.1 reference lines are drawn at the onset of rapid pressure rise in the carotid and at the dicrotic notch. These points on the carotid pressure curve nominally indicate the opening and closing of the aortic valve. The ECG and impedance
Fig. 3.1 - Esophageal Signals From a Dog
signal occur, on this time scale, with no delay. There is, however, a delay in the pressure signal, from its origin at the aortic valve to its reception at the transducer. This delay, composed of the delay in propagation of the pressure wave in the aorta, carotid, and catheter, could range from 10 ms to 100 ms depending on the propagation speed in the three tubes. In this thesis, for purposes of selecting a fiducial point the delay was assumed to be negligible.

The plateau just after the peak of the impedance signal was picked as the fiducial point indicating aortic valve opening. This point coincides directly with the start of rapid pressure rise in the carotid as recorded. If there were a 50 ms delay in the pressure recording, the peak of the impedance signal would be a better choice for the fiducial point corresponding to valve opening. Certainly more careful experimentation needs to be done to determine which point on the impedance curve best indicates valve opening, but as will be shown below, the parameters of the fiducial point detection routine can easily be changed to detect any point decided upon.

3.3 The Problem and its Solution

The problem then was to find PEP from the combination of the ECG and impedance signals as described above. From the definition of PEP, the start of the period is directly obtainable. The end of the period, however, must be found from a
fiducial point on the impedance signal. The fiducial point we derived and used was the plateau just after the peak as indicated by the arrow in fig. 3.1.

The ECG as measured from electrodes 1 and 10 (fig. 3.1) has a large p-wave and a tri-phasic QRS. It was found that for this data the most reliable point to trigger on was the large second slope of the QRS. The start of the Q-wave was then found by going backward through the stored data and looking for the first area of near zero slope. Figure 3.2 shows a sampled ECG with the start of the Q-wave as found by the program indicated.

The first idea considered for detection of the fiducial point on the impedance curve (which indicates the opening of the aortic valve) consisted of looking for the first sequence of positive and negative peaks in the second derivative following the absolute maximum in the impedance signal. One could then take the average in time of the two second derivative peaks as the point where the valve opens. This idea was abandoned before implementation for fear of the inherent noise in the derivatives, and the resultant likelihood of both false-positive and false-negative detections with this strategy.

What was needed then was a way of looking at a section of input data, comparing it with a model shape, and giving some indication of how close the match was. This suggested doing a correlation. It is a tenet of communications theory that a correlation receiver is optimal (9). This theory also implies that no signal correlates with a given signal better than itself,
Fig. 3.2 - Sampled ECG
and that the peak value of the correlation will occur when the two signals are lined up. One class of receivers, those that detect a pulse-position modulated signal, has similarities in problem definition to the application under consideration here. The primary difference is in the characterization of the noise. In communications, the noise is considered to be Gaussian random variable and by simply subtracting the energy of the input signal from the raw correlation, one may normalize inputs. For the purpose of detecting the fiducial point on the impedance signal, the noise, rather than being random, is of the same frequency as the signal of interest. It was found that normalizing the raw correlation by the a.c. rms amplitude of the input signal worked well and returned a number specifying how similar the input signal was to the model. This is analogous to normalizing the correlation of two variables by their standard deviation to produce a correlation coefficient.

For an input signal \( f(t) \) and a model signal \( g(t) \) the correlation integral is:

\[
R(\tau) = \int_{-\infty}^{\infty} f(t) g(t-\tau) dt
\]

If \( f(t) = g(t) \) then \( R(\tau) \) is the auto-correlation of \( g(t) \) and is maximum when \( \tau = 0 \). If \( f(t) \neq g(t) \), to compare \( f(t) \)'s correlation to \( g(t) \) with \( g(t) \)'s auto-correlation, one must normalize \( f(t) \) to
have the same rms value as \( g(t) \):

\[
R'(\tau) = R(\tau) \sqrt{\int_{-\infty}^{\infty} f^2(t) dt / \int_{-\infty}^{\infty} g^2(t) dt}
\]

For actual signal processing it is not possible to do an integral from minus infinity to plus infinity. In fact for the identification of a fiducial point on the impedance curve, finite windowing is preferable. Also inherent in signal processing in a digital computer is the quantification of time and amplitude of the signals. In this thesis a sampling rate of 500 Hz was used and the samples were quantized to a ten-bit or one part in 1024 accuracy. The actual equation used to compute the correlation for finite windowing and discrete time is:

\[
R''(N) = \frac{\sum_{k=1}^{33} f(N+k)g(k)}{\sqrt{\sum_{k=1}^{33} \left[ f(N+k) - \frac{1}{33} \sum_{k=1}^{33} f(N+k) \right]^2}}
\]

The term in the denominator is the energy of the input signal minus its mean in the window. If the energy of a signal over infinite time is finite then the mean is zero. In a finite window, however, there can be a non-zero mean which we do not wish to affect our calculations. Therefore, the mean is subtracted from the input signal before the energy is computed.

It was decided to use a 33 sample (66 ms) wide window. Using a model as shown in fig. 3.3 with the first slope, the plateau, and the second slope each 11 samples long, the
correlation curve in fig. 3.4 was produced.
The lower trace that extends the full width of the screen is 0.5 seconds of the sampled impedance signal. The shorter curve is
the value of the correlation as the model is moved along the
impedance curve (as is varied). As can be seen there is a
definite peak in the correlation at the plateau for which we are
searching. That is, there is a peak when the template best lines
up with the impedance signal. If the value of the peak is scaled
to the standard correlation coefficient, it has a value of 0.95
here.

One distinct advantage of using the correlation detector
routine described above is that if it is decided that another
fiducial point better represents the opening of the aortic valve,
then all that has to be changed is the 33 point template stored
in memory. If a detection routine had been written that relied
more on use of specific features of a specific shape of the
fiducial point rather than on a flexible model of it, major
revisions would have to be made and perhaps a totally new routine
written to detect the new point.

3.4 The Program

The algorithm developed to find PEP is best explained by
going through the sequence of execution of the program developed.
A flowchart of this program is given in fig. 3.5.

The program starts by sampling the ECG and impedance
Fig. 3.3 - Template used for Correlation Detection
Fig. 3.4 - Impedance Signal with Results of Correlation
Fig. 3.5 - Flowchart of PEP Program
waveforms at 500 Hz and storing the samples in a 256 sample circular buffer. When a QRS complex is detected by its steep down slope (the trigger point), 192 more samples are taken before processing starts. The first processing done is to go backward through the stored ECG data and find the start of the Q-wave. Once this is found, the time between it and the trigger point is computed and stored. Next the first peak of the impedance curve after the trigger point is found. If the impedance curve is decreasing at the trigger point, the trigger point is called the peak. Starting at this point, the first peak of the correlation greater than 0.90 is searched for. If no value of the correlation meets these conditions within 256 ms a "no-find" flag is raised. If, however, a peak is found, the time between it and the trigger point is computed and added to the stored Q-wave-trigger-point time and displayed as the PEP. In either case the program starts the sequence again at this point.

3.5 Program Testing and Performance

For testing the program was written to sample one heart beat, process the data, then present its results on the CRT if a peak of sufficient magnitude was found. Otherwise the program started over by sampling another heart beat. An example of this display is shown in fig. 3.6. The time between the arrows is the PEP.

Given that only data from one dog were available, it was
waveforms at 500 Hz and storing the samples in a 256 sample circular buffer. When a QRS complex is detected by its steep down slope (the trigger point), 192 more samples are taken before processing starts. The first processing done is to go backward through the stored ECG data and find the start of the Q-wave. Once this is found, the time between it and the trigger point is computed and stored. Next the first peak of the impedance curve after the trigger point is found. If the impedance curve is decreasing at the trigger point, the trigger point is called the peak. Starting at this point, the first peak of the correlation greater than 0.90 is searched for. If no value of the correlation meets these conditions within 256 ms a "no-find" flag is raised. If, however, a peak is found, the time between it and the trigger point is computed and added to the stored Q-wave-trigger-point time and displayed as the PEP. In either case the program starts the sequence again at this point.

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Given that only data from one dog were available, it was
Fig. 3.6 - ECG and Impedance with PEP Indicated
decided to test the program by varying the offset and gain of the signal applied to the A/D converter. The program was found to work well with input signals varying in size over a range of 8:1. Offset had no effect so long as the entire fiducial point was within the dynamic range of the A/D. When the program failed, it always was with false negatives (failure to find the PEP) and never with false positives (incorrect PEP values).

The PEP found for this dog was 80 ms. The random error was on the order of the sampling period (2 ms), which is the best that can be expected. This gives an error of ±3%. In humans the PEP is typically 100-150 ms, so the errors would be approximately the same for a human.
Chapter 4
CONCLUSION and DISCUSSION

4.1 Discussion

The approach taken in this thesis was empirical in nature: given the signals how does one find the points of interest. Although the algorithms developed using this approach work well for the data that we have recorded from one dog, a larger ensemble of signals should be analyzed.

From the position of the electrodes in the dog and from the way the impedance varied with the cardiac rhythm, decreasing during systole and increasing during diastole, the electrodes are probably sensing blood shifts in the aorta. This should be determined more directly, however, to determine whether the fiducial point being employed has a physiologic basis in the opening of the aortic valve. More careful experiments should also be done to determine the exact timing of the aortic pressure in relation to the impedance signal to find the most appropriate fiducial point to sense. Also, how does the impedance signal vary with changing cardiac function: in particular how does the shape of the fiducial point change. Perhaps the most important question along these lines is what does the impedance signal look like in the esophagus of a human, and does an appropriate fiducial point exist in such signals.

A significant question is whether more experiments should be done on dogs or whether the next step is to start getting data
from humans. For determination of the appropriate fiducial point
to use and to characterize the variations of it with different
physiological conditions, it makes sense to go directly to a
human studies program. This opinion is based on the significant
thoracic anatomical differences between humans and dogs. To
discover the exact origins of these impedance signals, primate
experiments would probably yield information more relevant to
humans than dog experiments.

Another point to be investigated is, given that the PEP is
being calculated with every heart beat or every other beat, how
can the value of PEP in previous beats be used to help find the
value on the next beat. Perhaps a form of processing such as
Kalman filtering can be productively applied here.

There appeared to be a high frequency (200 Hz) component to
the impedance signal at the point where the aortic valve opens.
Because of the bandwidth of the tape recorder used and the
sampling rate, it was difficult to decide whether the signal was
real or not. Hinohara reported measuring heart sounds directly
via an electrical impedance measurement from the esophagus (10).
We have not obtained a copy of this report as yet. If this high
frequency signal we are seeing is caused by the same fluid
motions that cause heart sounds, which in turn are associated
with valve movement, then that information might be useful in
determining systolic intervals more reliably.
4.2 Conclusion

In this thesis I have described systolic time intervals and how they could be measured as part of an esophageal probe monitoring system such as the one being developed in the Bioengineering unit. I have also described the methods used and the instrumentation used for recording the necessary ECG and impedance signals from an esophageal probe in order to develop an algorithm to measure the pre-ejection period. A brief description was given of the microprocessor development system and software tools used. The value of high-level languages was pointed out. I cannot over emphasize their value in a project such as this.

The development of the algorithm was based on a correlation method sometimes applied in communications theory and modified for this application. A program was implemented to find the PEP reliably from the data recorded from one dog, but several questions would have to be answered before this method of measuring PEP could be included as part of a practical esophageal monitoring system for use in humans.
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