Design of a Microprocessor-based Sphygmomanometer

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This paper describes the implementation on a microprocessor of a new method for the indirect measurement and recording of the systolic and diastolic blood pressure in humans. The technique is based on a statistical analysis of the cardiac pulse pressure signal. Polynomial relations are derived between the amplitude of the pulsatile pressure waveforms at the systolic and diastolic points and the amplitude of pulse signals detected when the artery is fully occluded. With the dual objective of automating the measurement procedure and minimizing errors, an electronic analog–digital sphygmomanometer that contains suitable electronic instrumentation was developed. The functions of processing the pressure signal, automating the measurement, and recording the results are performed and controlled by a microprocessor. A laboratory prototype embodying this approach was constructed and its performance and reliability were verified using a series of clinical tests. The test results indicate that the device is accurate within acceptable bounds for automated blood pressure instruments. (BIOMEDICAL INSTRUMENTATION & TECHNOLOGY 1990;24:31–36)

In recent years, the importance of reliable measurements of systolic and diastolic blood pressures in humans has been recognized. Available measurement techniques are generally classified into two categories: direct and indirect.[1,2] Direct or invasive techniques are more accurate and reliable, but require expert medical intervention and may be the cause of patient discomfort. Indirect methods rely on the use of an inflatable occlusive cuff followed by analysis of the Korotkoff sounds by either stethoscopic or electronic auscultation methods.[3-5] Other methods have also been proposed; these are generally based on the detection and measurement of pressure signal characteristics.[6-8]

Automated blood pressure techniques are generally distinguished into those that employ a microphone to detect the onset and termination of Korotkoff sounds and those that identify the magnitude and location of the systolic and diastolic points on the basis of a priori information regarding their characteristics. Unfortunately, devices based on these techniques do not yield reliable results when used by novice operators. Experience has shown that their utility requires the intervention of clinical personnel. It is obvious, therefore, that a new indirect method is needed that allows for the reliable automated measurement and recording of the systolic and diastolic pressures in humans. The cost-effective implementation of any new technique must take into consideration the latest advances in microcomputer technology, which facilitates considerably the design, development, and control of the measurement procedure.

Analysis of a statistical sample of clinical data has shown a consistent relationship between the amplitude of the "background" pulse signal that is detected with a sensitive pressure transducer when the artery is fully occluded and the corresponding amplitudes of the systolic and diastolic pressure points. Such a relationship is conveniently described in terms of polynomials. The coefficients of these polynomials are derived via curve fitting of the clinical data. In accordance with these findings, the measurement procedure is initiated by detecting and storing the amplitudes of the background pulses. The expected amplitudes of the systolic and diastolic points are computed next through the polynomial relationships.

Electronic blood pressure instruments that exploit either the Korotkoff sounds or some other detection scheme are of the analog, digital, or hybrid variety. The design philosophy is basically the same for all of them: pressure transducer, electronic processing of the signal, display. They differ as to the principle of operation. Thus, proposed devices have used the minimax peak detection technique,[9] strain gauges,[10] analog preprocessing,[11] balanced blood pressure amplification,[12] and numerical display.[13]

This paper describes the development of an electronic sphygmomanometer based on the observations referred

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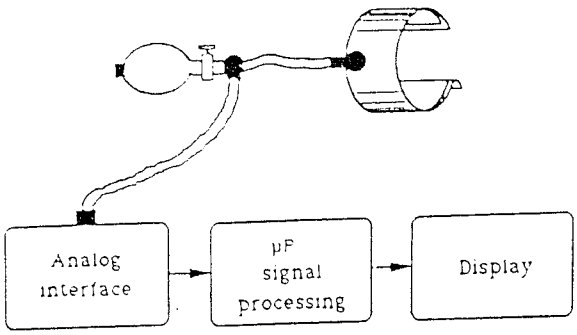

Figure 1. Simplified schematic of the main parts of the proposed sphygmomanometer.

to above and described in detail elsewhere.[14, 15] A prototype instrument was constructed using a microprocessor with appropriate analog interfaces. The advantages offered by the microprocessor in processing the pressure waveforms contribute to the improved measurement accuracy and robustness of the device in its ability to reject noise and identify false measurements. The microprocessor also provides the ability to store test results for further analysis and diagnostic purposes.

GENERAL DESCRIPTION

Figure 1 shows the general functional diagram of the proposed instrument. A conventional occluding cuff with a mechanical relief valve is retained. The cuff pressure mechanism may be easily automated by means of an appropriate air pump regulated by the microprocessor. Cuff pressure is transferred to a pressure transducer whose electrical output is proportional to absolute pressure in the brachial area. The transducer output signal, after appropriate preprocessing, branches into two paths and is directed to the input of the microprocessor. Two analog-to-digital (A/D) converters are utilized to convert an analog signal to digital form for further processing by the microprocessor. The signals fed to the two branches are subjected to different processing techniques. At each sampling period the first branch computes the mean value of the pressure waveform (P), while the second filters out the mean value of the signal, thus leaving only the oscillating component that represents cardiac pulses. At the end of the measurement cycle, the microprocessor has estimated the systolic (SP) and diastolic (DP) cardiac pressures, which are appropriately displayed. The measurement procedure can be understood with the assistance of Figure 2, which depicts the electrical signal p(t) at the output of the pressure transducer as a function of time t. The figure is not drawn to scale, in order to emphasize the signal characteristics. The same figure shows the amplitudes (S,D) of the oscillations during the systolic and diastolic time instants, as well as the values of SP and DP. The velocity of cuff pressure deflation influences measurement accuracy. The latter is inversely proportional to the speed of cuff deflation. An acceptable accuracy level is achieved by setting the rate of cuff pressure deflation at a value below a certain limit. The amplitudes of the systolic and diastolic pulses are related to the amplitude of the background pulses,[14] as shown in Figure 2, via the relations:

$$S = B \times SR \quad (1)$$

$$D = B \times DR \quad (2)$$

The quantities SR and DR, which refer to the systolic and diastolic ratios, respectively, are not constant, but depend upon the amplitude of the background pulses, as demonstrated by statistical analysis of clinical data. Thus, a correct cardiac pressure measurement requires the implementation of some type of an adaptive control algorithm. The dependence of SR and DR on the background pulse amplitude is derived from a statistical analysis of a sufficient sample of clinical data obtained with an early version of the instrument. This functional dependence is approximated fairly accurately by two polynomial relations of the form:

$$SR = s_5B^5 + s_4B^4 + s_3B^3 + s_2B^2 + s_1B + s_0 \quad (3)$$

$$DR = d_5B^5 + d_4B^4 + d_3B^3 + d_2B^2 + d_1B + d_0 \quad (4)$$

Furthermore, the statistical processing has indicated that increased accuracy is achieved when the initial pressure level (P_o) is set according to the expected region of the systolic pressure of the individual under test. Three different initial values for the pressure P_o are, therefore, used: 160 mmHg, 200 mmHg, and 240 mmHg. The above two adaptive procedures, when combined, result in three

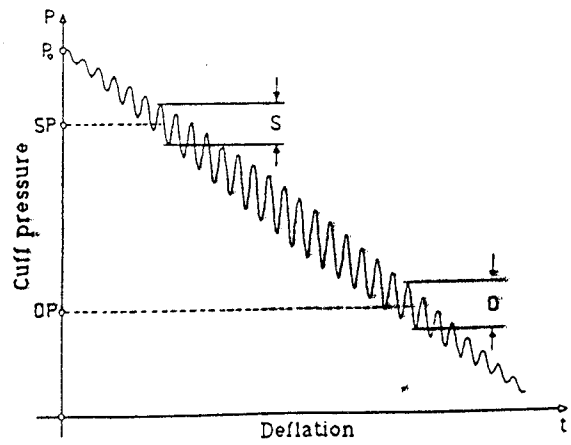

Figure 2. Schematic of the combined signal from the pressure transducer during cuff deflation.

Table 1. Coefficients s_n and d_n of the Polynomials (see text)

Coefficient	P_o = 160 mmHg	P_o = 200 mmHg	P_o = 240 mmHg
s_5	0.0	0.66998488	0.0
s_4	−0.14562118	−7.242701	1.1758647
s_3	0.47420055	29.495819	−14.9217
s_2	2.2125144	−55.41825	40.5957
s_1	−10.92095852	46.44429	−44.86037
s_0	15.20197582	−9.249882	21.55773
d_5	0.06916247	1.07702947	0.0
d_4	−0.758686	−10.711811	−18.7053966
d_3	2.20527744	39.74743652	85.5096664
d_2	2.27368307	−68.038574	−135.8229523
d_1	−19.6591301	52.40162	85.2866897
d_0	27.327362	−9.1540079	−10.71791

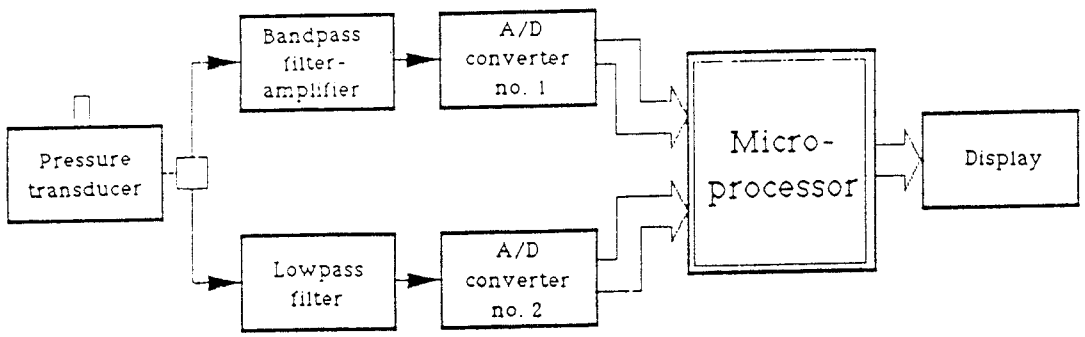

Figure 3. Block diagram of the analog and digital sections of the sphygmomanometer.

groups of relations of the form (3) and (4), one for each value of the initial pressure. The coefficients s_n and d_n of the polynomials (3) and (4), as determined from the statistical analysis of clinical data, are given in Table 1.

The following steps outline the measurement procedure:

1. The instrument is set at one of the initial values of P_o, which is larger than the expected value of the systolic pressure by at least 40 mmHg.

2. The cuff is occluded until the cuff pressure reaches the value P_o, at which point the instrument flashes a warning signal.

3. Cuff deflation begins at this stage, and the instrument monitors continuously the value of the cuff pressure (P), as well as the pressure waveform whose amplitude is computed at each sampling interval.

4. The mean value of the background pulse amplitude is estimated next from a sample of the first five pulses. This mean is taken as the amplitude of the backgound pulses (B).

5. The systolic (SR) and diastolic (DR) ratios are calculated through relations (3) and (4) and taking into consideration the amplitude B.

6. The amplitude of the systolic (S) and diastolic (D) pulses is computed through equations (1) and (2) and the amplitude B.

7. The time instant at which the amplitude of the oscilating pressure waveform becomes equal to the value S is detected. At this time instant, the value of the cuff pressure, which corresponds to the systolic pressure (SP), is simultaneously stored and displayed.

8. Finally, the time instant at which the waveform amplitude is equal to D is detected. This point corresponds to the diastolic pressure (DP), which is simultaneously stored and displayed.

DESIGN METHOD

The procedure described above is programmed to be executed by a microprocessor, which is the main component of the proposed instrument. A block diagram of the overall configuration (Fig. 3) clearly distinguishes the parallel paths followed by the electrical signal after it exits the pressure transducer. This signal separation is deemed necessary since the pressure waveform is but a small per-

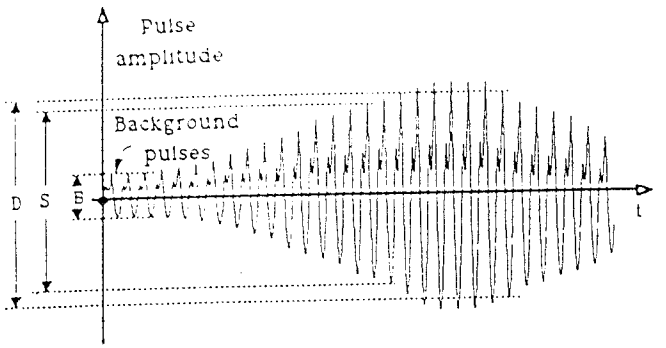

Figure 4. Cardiac pressure oscillations extracted from the combined signal of Figure 2.

centage of the whole pressure signal. Otherwise, for satisfactory accuracy of the quantities referred to the pressure waveform (instantaneous value of waveform, background value), a high-resolution A/D conversion would be required. Such an implementation would reduce processing speed, however, and as a consequence other problems such as increased cost of the instrumentation would arise.

One part of the signal is fed to a bandpass filter, which has a bandwidth from 0.5 to 2.5 Hz. That is, it permits the passage of those cardiac pulses in the range from 30 to 150 pulses/min, while it cuts off completely the steady-state component (mean pressure value) and the higher harmonics that are due primarily to respiratory effects and arm movements of the subject. The same stage performs an appropriate amplification of the signal so that the resolution of the A/D converter is fully utilized. Well-distinguished cardiac pulses appear at the output of this stage ready to be fed to the A/D-1 of the microprocessor.

The second part of the signal is fed to a low-pass filter so that undesirable higher harmonics may be rejected. This stage performs a triple function: 1) it amplifies the signal so that the latter fully utilizes the region of the input voltage to the A/D converter; 2) it functions as a low-pass filter for the further rejection of higher harmonics; 3) the constant offset voltage superimposed on the signal by the pressure transducer is eliminated through the assistance of a potentiometer. The output signal from this stage is fed directly to the A/D-2 input of the microprocessor and is proportional to the mean value of the cardiac pressure.

Software

After program initialization, the initial pressure value P_o, that is, one of the set values of 160, 200, or 240 mmHg (step 1 of the measurement procedure described above), is input. The P_o value is used, on one hand, to provide a warning signal when the actual cuff pressure reaches P_o during inflation. On the other hand, a particular set of the polynomial relations (Table 1) is selected on the basis of the P_o value.

Subsequently, the output of the A/D-2 converter, that is, the cuff pressure, is monitored during initiation of cuff inflation (step 2). At each sampling time, the instrument checks whether the cuff pressure has reached the preset value of P_o. When this condition is reached, a signal warns the operator to terminate cuff inflation. From this point on, cuff deflation begins, with a constant and appropriately regulated speed (step 3). Now the instrument monitors continuously the oscillating component of the pressure via the A/D-1 (step 3). The instantaneous value P of the waveform is observed at the connecting port of the A/D-1 converter and a check is performed to identify whether P has reached a maximum or a minimum. When P reaches an extremum, the following process is executed.

First, the time instant of a maximum or a minimum, and its value, are recorded. This value is subtracted from the previous corresponding extremum and the amplitude of the waveform is computed and stored in an appropriate array (step 3).

Next, when the second cardiac pulse appears, the cardiac pulse rate is computed and displayed in pulses min. If the pulse order is less than 5, then tracking of the A/D-1 port is repeated. When the fifth cardiac pulse is detected, the following procedural steps are executed:

1. The mean value of the amplitudes of the first five pulses is computed. This value is stored as the amplitude B of the background pulse (step 4). This step is performed in order to reduce the effects of the transient phenomena at the beginning of the measurement procedure. Here, the adaptive control procedure of the proposed measurement method takes place.

2. The values of the systolic (SR) and diastolic (DR) ratios are computed, taking into account the selected polynomial set and the amplitude B of the background pulse (step 5).

3. With the known values of SR, DR, and B, the amplitudes of the systolic (S) and diastolic (D) pulses are computed adaptively (step 6). At this point, the expected amplitudes of the systolic and diastolic pulses are known.

From the sixth pulse on, the following takes place:

1. The pulse amplitude is checked to see whether it is equal to S. In the event that this is true, the value P of the cuff pressure is read by the A/D-2 converter. This last value is the systolic pressure (SP), which is stored and displayed (step 7).

2. The procedure continues by monitoring the pulse amplitude. When this becomes equal to D, the value of the cuff pressure is read by the A/D-2 port and this value corresponds to the diastolic blood pressure (DP). This DP value is stored and displayed (step 8). With this step, the measurement procedure terminates. The cardiac pulse rate

and the systolic and diastolic pressures are available for displaying or further processing.

3. If the events of steps 1 and 2 above are not observed, then monitoring of the A/D-1 port as well as the procedural steps 1 to 2 are repeated.

RESULTS

Experimental results for verification of the proposed blood pressure measurement method were derived using a laboratory-constructed prototype. To keep observer errors to a minimum level, clinical tests were conducted by a large number of trained investigators. Parallel measurements were also taken using a conventional sphygmomanometer so that the accuracy and reliability of the proposed method could be assessed by comparison with the conventional method.

Figure 5 shows typical samples of comparative measurements using the proposed and conventional methods. A comparison of systolic pressure measurements is exemplified in Figure 5*A*, which shows both the conventional systolic pressure and the device's systolic pressure. The largest deviation found between the two techniques was 6 mmHg. Figure 5*B* shows a similar comparison for diastolic pressure. The largest deviation between the two methods was 5 mmHg. Thus, the accuracy of the device

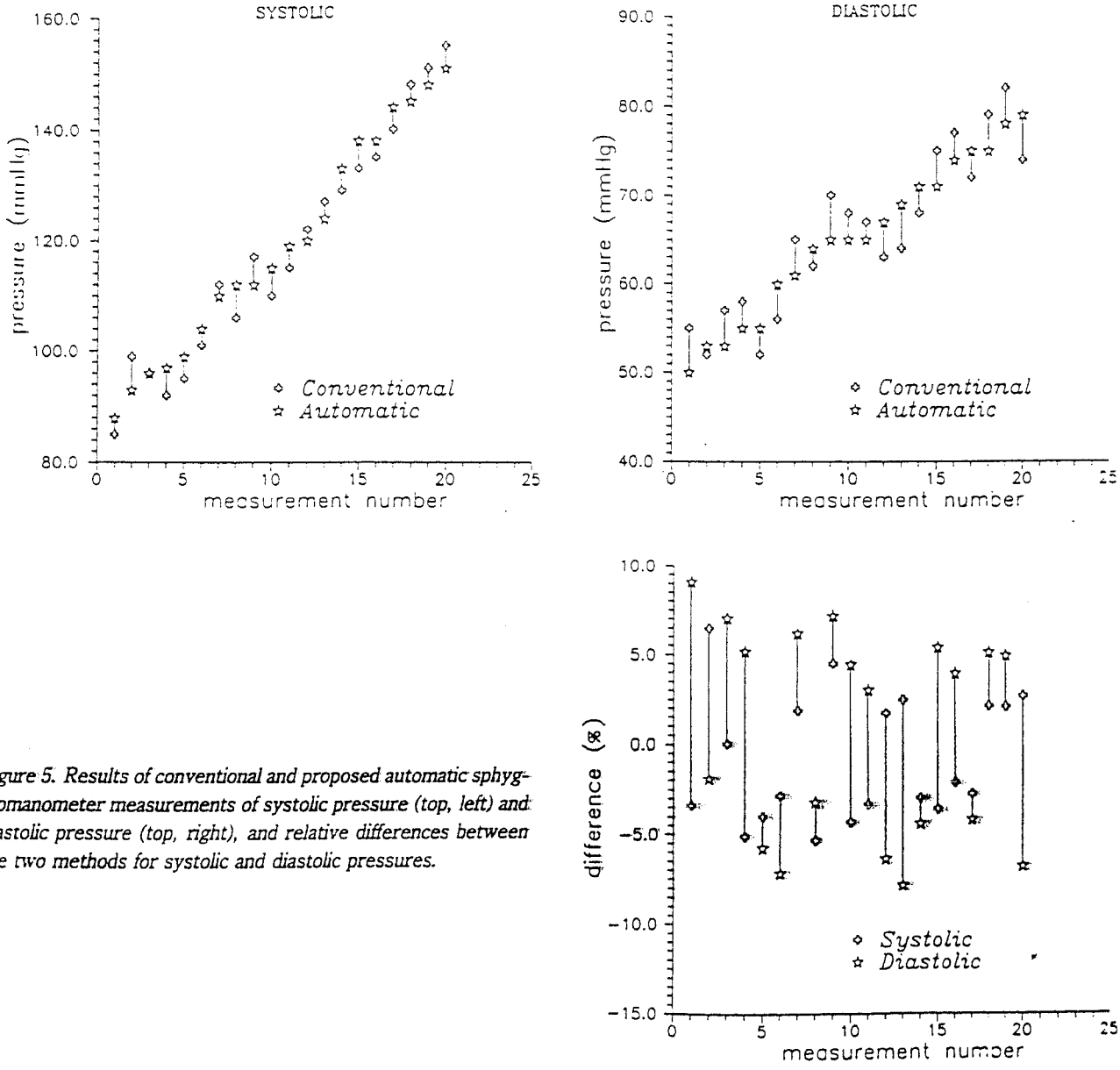

Figure 5. Results of conventional and proposed automatic sphygmomanometer measurements of systolic pressure (top, left) and diastolic pressure (top, right), and relative differences between the two methods for systolic and diastolic pressures.

lies well within accepted limits according to AAMI standards.

The percentage deviations with the conventional method as a base are shown in Figure 5C. The maximum deviation in the measurements of systolic pressure was ±6%, while that for diastolic pressure was within ±7%. In Figure 5, each of the data points shown is actually the mean value resulting from a number of test measurements on the same individual.

The most significant sources of error are estimated to be 1) motion of the subject's arm, 2) the respiratory function of the individual, and 3) any severe cardiac ailment. A smaller percentage error (approximately 0.5%) is attributed to the device itself because of transducer nonlinearities and the accuracy associated with the A/D converters. The observer error in interpreting the results is negligible since blood pressure readings are recorded digitally.

CONCLUSIONS

The proposed device may be developed on an autonomous basis using a dedicated microprocessor. The resulting implementation is relatively inexpensive and provides flexibility and ease of operation. It may also be integrated into an existing computer system, such as a PC equipped with a two-channel A/D board, with provision for outside processing of analog signals (pressure transducer, filters, amplifiers, and so on). The measurement program is loaded only when a cardiac pressure measurement is to be performed. The computer system is available for other functions during the remaining time.

The proposed blood pressure measurement method, which is based upon the relationship between the background pulses detected when the artery is fully occluded and the magnitudes of the systolic and diastolic pressures of an individual, as implemented on a microprocesser, performs satisfactorily, as indicated by clinical tests. Assessment of the device's performance was based on comparative results obtained from a conventional sphygmomanometer. A more accurate verification procedure would involve comparison with results of direct (invasive) techniques, but, such a comparison is difficult to implement, for obvious reasons. AAMI (1982) standards for electronic or automated sphygmomanometers suggest the auscultatory method for comparison purposes, since it is easier to implement than are invasive techniques. ■

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