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Development of a microprocessor-based adaptive technique for blood pressure measurements

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Summary

The paper introduces a new microprocessor-based adaptive technique for the indirect measurement of the systolic and diastolic pressure in humans. The technique is based upon a statistically consistent relationship between the amplitude of the pulsative pressure waveform at the systolic and diastolic points and the amplitude of pulse signals detected when the artery is fully occluded. An adaptive measurement philosophy has been implemented in the design of an electronic analog-digital sphygmomanometer which, in addition to a pressure transducer, contains suitable electronic instrumentation for processing and displaying the electronic signals. A dedicated microprocessor is used to store statistical relations and control the operation of the device. Verification of overall system accuracy is accomplished via direct comparison with manual auscultatory measurements. Clinical testing of a prototype indicates satisfactory performance; measurement errors are maintained well within proposed standards for automated sphygmomanometers.

Introduction

Accurate clinical blood pressure measurements have been a concern ever since the indirect method was developed at the turn of the century [5, 6]. Conventional noninvasive methods for blood pressure measurement rely on the use of an inflatable occlusive cuff followed by analysis of the Korotkoff sounds by either stethoscopic or electronic auscultation methods [3, 7, 12]. Other available techniques are based upon the oscillometric method [15], or measure mean arterial pressure [10], or long-term variations of blood pressure by peak-and-trough detection of the pressure waveform [8].

Analog, digital, and hybrid electronic techniques have been utilized for the detection and recording of the systolic and diastolic blood pressure in humans. Some automatic devices not using Korotkoff sounds, e.g., the 'Dinamap' machine, have reached the stage of commercial exploitation. Strain gauge or pressure transducing devices are usually coupled to signal processing instrumentation and display equipment. Minimax peak detecting electronic apparatus for identifying the systolic and diastolic pressure levels have been proposed [2]. Strain gauge techniques [4], analog preprocessing [11], appropriate blood pressure amplifier means [12], and display devices [9] have been de-

veloped and reported in the literature.

In principle, the auscultatory techniques are based upon the inclusion of a microphone in the cuff area for the direct detection of the Korotkoff sounds or via comparison methods with a reference signal or through identification of peak pressure amplitudes in an 'open loop' configuration. These basic design concepts require human intervention in recording the appropriate pressure levels. They are usually characterized by the need for medical supervision during measurement and a lack of performance reliability. The measurement complexity and lack of reliability have prevented the widespread nonprofessional use of these devices despite a growing awareness of hypertension as a serious health hazard and an increasing recognition of the importance of early detection and treatment of the disease.

The authors have reported a new technique for the measurement of the systolic and diastolic blood pressure in humans [13]. The technique is based upon a statistically consistent relationship between the amplitude of the pulsative pressure waveform at the systolic and diastolic points and the amplitude of the pulse signals detected when the artery is fully occluded. Clinical evidence has substantiated this proportional relationship between the amplitude of the artery occluded pulses (background amplitude) and those resulting during deflation of the cuff and corresponding to the onset and termination of the Korotkoff sounds [14]. The relationship was found to be independent of the individual tested. These relationships have been exploited in the design of an analog electronic sphygmomanometer, which, in addition to the pressure transducing device, contains suitable electronic instrumentation for processing and displaying the systolic and diastolic signals.

Statistical analysis of the clinical data has also shown that the values of the systolic and diastolic ratios for each region, at which the amplitude of the background pulses is measured, are functions of the background pulse amplitude. This functional dependence may be approximated by a polynomial of a suitable order. It is sufficient, therefore, to measure the background pulse amplitude of an individual and capitalize upon this functional dependence in order to determine the cardiac pressure ratios. This approach requires a microprocessor (μP) device for its implementation. The development of such a μP-based adaptive technique for cardiac pressure measurement is reported in this paper. The resulting instrument is characterized by a higher degree of accuracy and reliability. Accuracy of performance is well within acceptable standards when compared with conventional means.

General description

Figure 1 shows the general functional diagram of the proposed instrument. A conventional occluding cuff with a mechanical relief valve is retained in the scheme. The cuff pressure mechanism may be easily automated though via 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 breachial area. The transducer output signal, after appropriate preprocessing, branches into two paths and is directed to the input of the microprocessor. Two A/D converters are utilized to convert an analog signal to digital form for further processing by the microprocessor (μP). The signals fed to the two branches are subjected to different processing techniques. The first branch computes at each sampling period 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 due to cardiac pulses. At the end of the measurement cycle, the μP has estimated the systolic (SP) and diastolic (DP) cardiac pressures which are appropriately displayed. The measurement procedure can be understood with the assistance of Fig. 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 value of the systolic (SP) and diastolic (DP) pressures. The velocity of cuff pressure deflation influences the measurement accuracy. The latter is inversely pro-

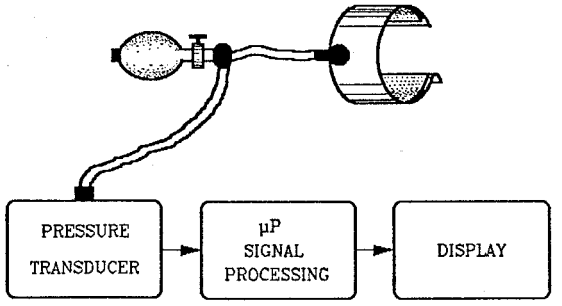


Fig. 1. Simplified schematic of the main parts of the proposed sphygmomanometer.

portional 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. According to the analysis reported in [13], the amplitude of the systolic and diastolic pulses is related to the amplitude of the background pulses, as shown in Fig. 2, via the relations:

$$S = B \times SR \tag{1}$$

$$D = B \times DR \tag{2}$$

The quantities SR and DR, referred to as the systolic and diastolic ratios, respectively, are not constant, but they themselves depend upon the amplitude of the background pulses (B), as asserted by statistical analysis of clinical data. Thus, a correct blood 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:

$$SR = s_5B^5 + s_4B^4 + s_3B^3 + s_2B^2 + s_1B + s_0 \tag{3}$$

$$DR = d_5B^5 + d_4B^4 + d_3B^3 + d_2B^2 + d_1B + d_0 \tag{4}$$

Furthermore, the statistical processing has indicat-

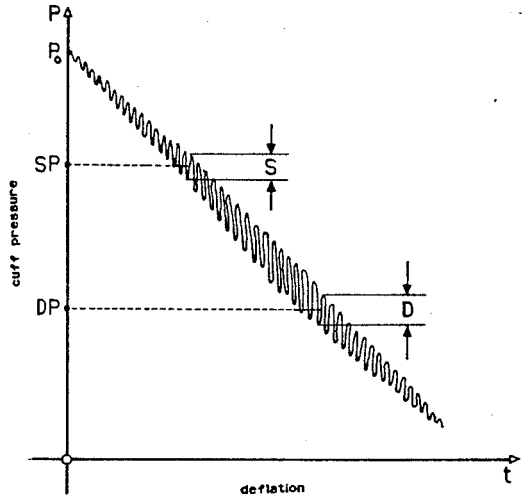


Fig. 2. Schematic of the combined signal from the pressure transducer during cuff deflation.

ed 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, 240 mmHg. The above two adaptive procedures, when combined together, result in three 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.

Large differences in the coefficients for different P_o values (although the polynomials have similar shapes) arise because of the following two reasons: (1) A small movement of the peak of the waveform, as dictated by the measurement set and fitted with a fifth degree polynomial, may result in this peak leaving the polynomial's extremum and approaching its neighbor polynomial, causing, therefore, a drastic change in the values of the coefficients. (2) The set of points (x, y) for each waveform that are used to construct the corresponding polynomial do not have the same x values in each case; we may, therefore, expect coefficients of different values. Furthermore, these coefficients are not totally independent of individual properties, such as body weight, age, tissue composition, etc., and for dif-

ferent diseases of the subjects tested (e.g. cardiac valve diseases, vascular diseases, etc.). We reject presently from the measurement set extreme pathological conditions that may lead to large parameter variations. Their variability and means to account for it are the subject of current investigations.

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 5 pulses. This mean is taken as the amplitude of the background 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 oscillating 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

Hardware description

The procedure described above is programmed to be executed by a μP, which is the main component of the proposed instrument. The block diagram of the overall configuration is shown in Fig. 3. The figure 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 percentage of the whole pressure signal. Otherwise, for a satisfactory accuracy of the quantities referred to the pressure waveform (instantaneous value of waveform, background value) a high resolution analog to digital conversion would be required. Such an implementation though would result in reduced processing speeds and as a

Table 1. Coefficients for the approximation of SR and DR (eqs. 3 and 4).

Coeff.	P_o = 160 mm Hg	P_o = 200 mm Hg	P_o = 240 mm Hg
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	47.44429	− 44.86037
s_0	15.20197582	− 9.249882	21.55773
d_5	0.06916247	1.07702947	0.0
d_4	− 0.758686	− 10.711811	− 189.7053966
d_3	2.20527744	39.74743652	85.5096664
d_2	2.27368307	− 68.03857	− 135.8229523
d_1	− 19.6591301	52.40162	− 85.2866897
d_0	27.327362	− 9.1540079	− 10.71791

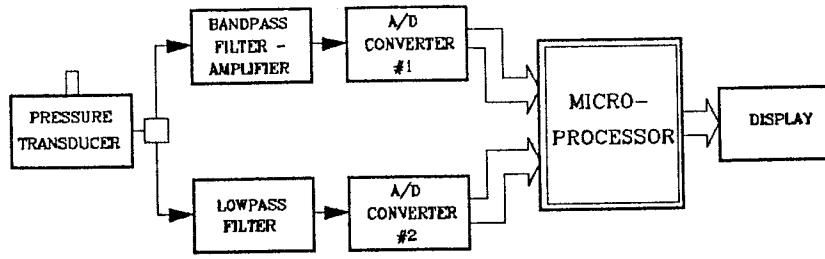


Fig. 3. Block diagram of the analog and digital sections of the sphygmomanometer.

consequence other problems such as increased cost of the instrumentation would arise.

One part of the siginal is fed to a bandpass filter, where the quasi steady state component of the pressure (the value of the pressure P) is eliminated. The same filter also cuts off higher harmonics which do not contain useful information for the present application. As a result, the only component at the filter output is the oscillating part of the pressure waveform which is subsequently amplified to take the form shown in Fig. 4. This last part of the signal is fed to the input of the first converter, A/D-1, with a 12 bit resolution and a conversion time of 1 ms. This resolution is considered to be satisfactory since it can monitor $2^{12} = 2096$ levels of the signal, i.e., a conversion accuracy of about 0.024%. The conversion time is also considered satisfactory since, for a typical cardiac pulse rate of 80 pulse/min, the resulting cardiac pulse period is equal to 750 ms. With a conversion time of 1 ms, therefore, 750 samples of each pulse are taken during each period. The output of the A/D becomes the input to one of the μP input ports.

The second part of the signal is fed to a low pass filter so that undesirable higher harmonics may be rejected. The filter output is taken to a second A/D converter (A/D-2) with similar technical characteristics as the previous one. The digital output of this converter corresponds to the value of the cuff pressure (P) and is fed to another port of the μP.

The main signal processing task is undertaken by the μP. The measurement results are displayed on a suitable display device connected to the μP; the display device could also be a videoscreen.

Figure 5 depicts the detailed electronic diagram of the analog section. The pressure transducer is a linear hybrid integrated circuit of the type LX1602G, manufactured by National Semiconductors. This device incorporates a piezoresistive sensor as part of a balanced bridge arrangement. Cuff pressure through a plastic tubing is transferred to the transducer input while an analog voltage appears at the output port. The transducer calibrated range is from 0 to 776 mmHg and the conversion coefficient is 13 mV/mmHg with an accuracy of ± 0.25 mV/mmHg (1 mmHg = 0.0193 psi). The maximum calibration error is $\pm 1.5\%$ of full span. The output signal contains also an offset voltage of 2.5 V (± 0.3 V). The electrical signal (proportional to the blood pressure) is fed from the pressure transducer output to a low pass filter stage which is designed around an LF356 operational amplifier with FET inputs. Subsequently, the signal is fed to the next stage which is also based on an LF356 IC. This stage performs a triple function:

1. It amplifies the signal so that the latter utilizes fully 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 μP and is proportional to the mean value of the cardiac pressure.

The output signal from the previous stage is also directed to the next stage organized around an LF356. This stage has been designed as a band pass filter with a bandwidth from 0.5 to 2.5 Hz. That is,

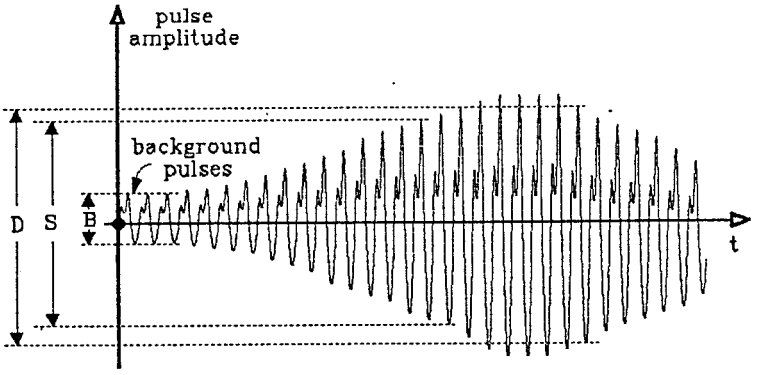


Fig. 4. Cardiac pressure oscillations extracted from the combined signal of Fig. 2.

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 which 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 μP.

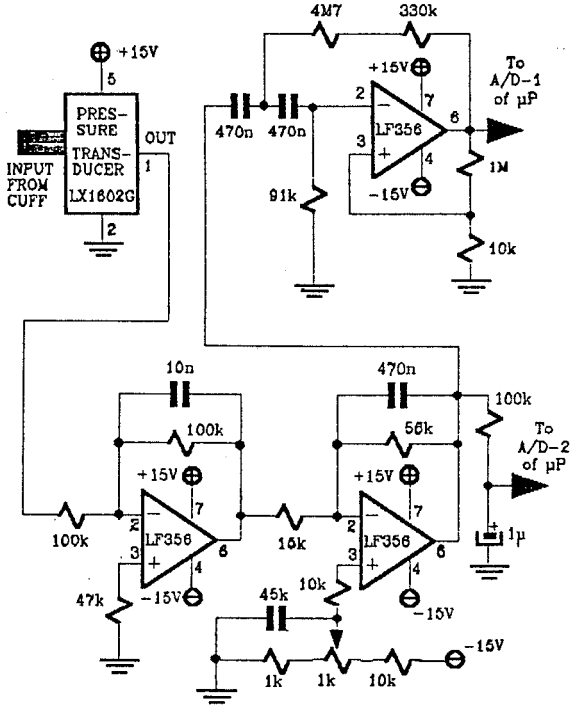


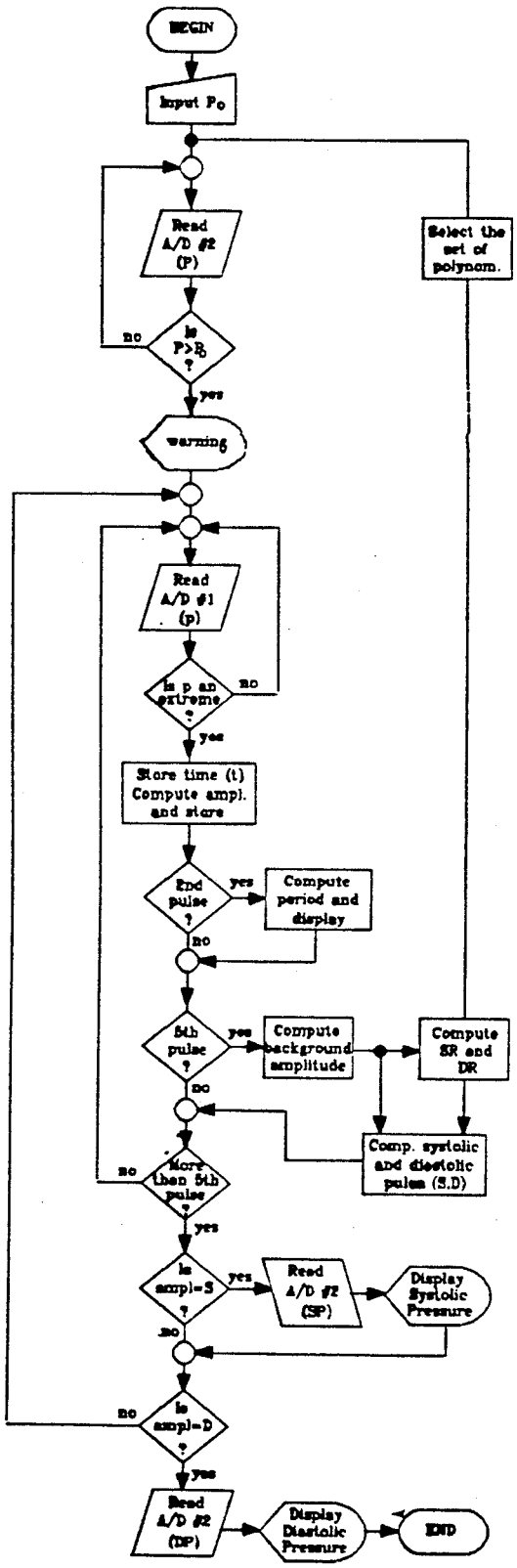
Fig. 5. Comparison of the conventional and the proposed measurement results.

Software description

The detailed measurement task is allocated to a suitable computer program which may be written in any of the conventional programming languages (assembler, basic, pascal, C, etc.). Figure 6 shows the flow diagram of the program used in the proposed instrument. We detail below the operating characteristics of the instrument's software package with the help of the flow diagram.

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

Subsequently, the output of the A/D-2 converter is monitored, i.e., the cuff pressure, while cuff inflation has been initiated (step 2). At each sampling time, the instrument checks if 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 the value of P has reached a maximum or a minimum. When the pressure P reaches an extremum, the following process is executed.

First, the time instant of a maximum or a minimum, as well as 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 though, the following procedural steps are executed:

1. The mean value of the amplitudes of the first 5 pulse is computed. This value is stored as the amplitude B of the back ground pulse (step 4). This step is performed in order to reduce the effect 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 6th pulse on, the following takes place:

Fig. 6. Flow diagram of the program for blood pressure determination.

1. The pulse amplitude is checked to see if 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 values 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, as well as the systolic and diastolic pressure levels 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.

The block diagram of Figure 6 does not contain a number of details for purposes of preserving the clarity of presentation. For example, it does not depict the part of the program that is concerned with the detection of local minima and their subsequent rejection, thus avoiding severe measurement problems. Care has also been taken to reject sharp changes in the values of the extrema which are due to sudden arm movements or the respiratory functions of the subject.

Results

A prototype sphygmomanometer was constructed and tested with a conventional mercury manometer taken as the standard. An attempt was made to use commercially available (of the inexpensive type) automatic devices in the evaluation process. The performance of these machines, however, was erratic and the attempt was abandoned in favor of the conventional mercury manometer as recommended by AAMI.

Clinical test studies were used to verify the reliability and accuracy of the proposed method. In parallel, similar results were recorded using conventional means for comparison purposes. Multiple observers were used for both the conventional and the proposed technique in order to minimize errors due to the observer.

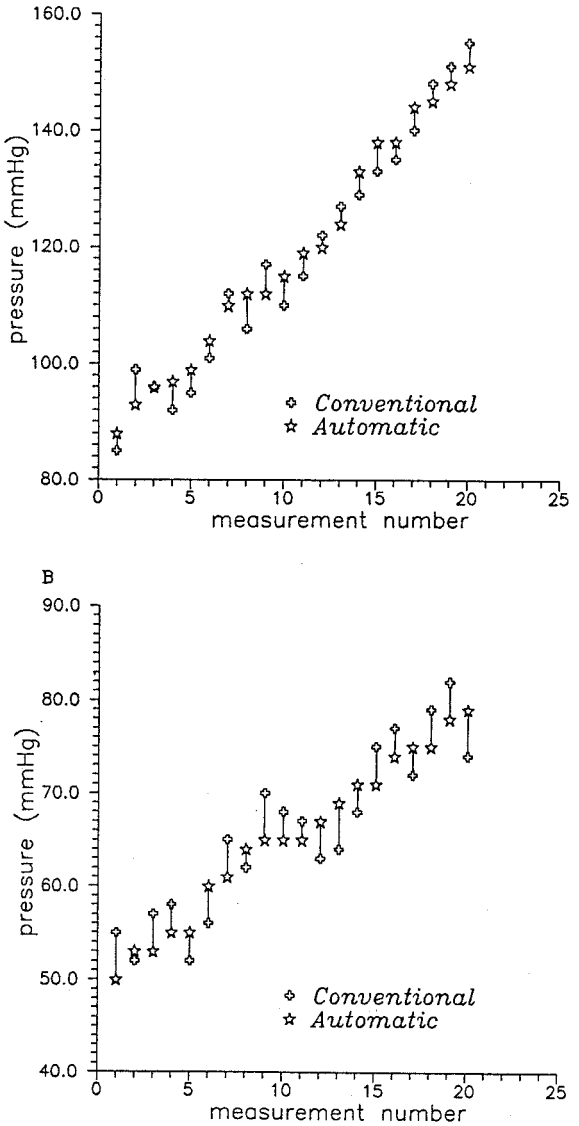


Fig. 7A–B. Comparisong of measuring methods for the systolic pressure.

Figures 7a, 7b, and 7c show a typical sample of comparative measurements using the proposed and conventional methods. A comparison of the systolic pressure measurements is depicted in Figure 7a. Both the conventional systolic pressure and the device systolic pressure are shown in the figure.

The largest deviation between the two tech-

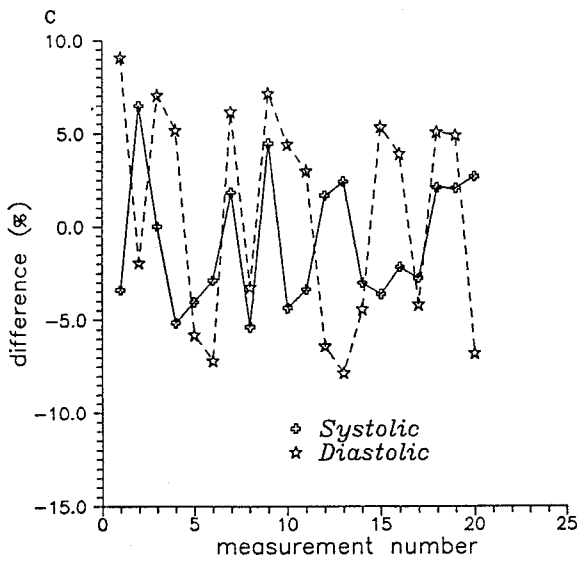


Fig. 7c. Relative difference between the two melthods.

niques is measured to be ±6 mmHg. Figure 7b shows a similar comparison for the diastolic pressure. Again, the largest deviation between the two methods is 5 mmHg. The accuracy of the device lies well within accepted limits according to AAMI standards. The percent deviation with the conventional method as a base is shown in Figure 7c. The maximum deviation in the measurement of the systolic pressure is ±6%, while the deviation for the diastolic pressure is found to be within ±7%. In the above figures, each of the data points shown is actually the mean value resulting from a number of test measurements on the same individual.

Possible error sources may be due to:

1. The observer
2. The equipment
3. The subject tested
4. The environmental conditions.

The proposed instrument is designed to operate with a minimum observer involvement, thus minimizing the error contribution from this category. Errors due to equipment arise primarily from the nonlinear characteristics of the electronic pressure transducer (of the order of 1%) and the other components (amounting to approximately 0.5% of full scale). The subject tested may contribute a substantial part of the experimental error by allowing motion of the arm during measurement, abnormal breathing, etc. Finally, environmental conditions, such as vibrations, noise, etc., may affect the accuracy of the measurement.

Conclusions

The proposed device may be developed on an autonomous basis using a dedicated μP. 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, etc., as shown in Fig. 5). 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.

A theoretical verification of the proposed technique's system efficacy is difficult to establish, if not impossible, owing to the character of the systolic and diastolic pressure levels, the inherent inaccuracies, etc. It is generally accepted, therefore, that overall system accuracy may be verified by comparison with conventional (auscultatory and invasive) techniques (AAMI, 1982). Care must be exercised in organizing and conducting the clinical testing and analyzing the statistical data so that acceptable system accuracies are assured. The invasive method of verifying noninvasive blood pressure measurement devices is potentially the most precise means of verification. At the same time, it is practically more difficult to obtain subjects and conduct measurements. The AAMI standard for electronic or automated sphygmomanometer concludes that, 'of the two methods, the first (auscultatory) is by far the easiest to implement and the generally acceptable'.

The proposed sphygmomanometer performs satisfactorily when compared with conventional standards. The design methodology adopted is based on a statistically consistent relationship be-

tween the pulse amplitude detected with the brachial artery fully occluded and the systolic and diastolic cardiac pressure points of an individual. The basic principle has been validated through extensive clinical testing.

The instrument design philosophy incorporates an analog-digital approach which leads to a reliable device realization within acceptable accuracy limits and reasonable construction costs. Special attention has been given to the design of appropriate signal processing circuits so that errors due to environmental conditions or random arm movements are minimized and the desired signal is clearly distinguished from noise levels.

Commercial oscillometric devices, such as the DINAMAP, depend for their operation on the fact that maximal oscillations of cuff pressure occur when cuff pressure is equal to the mean brachial artery pressure. The method proposed in this paper is based on a different concept-adaptive comparison of systolic and diastolic levels to the amplitude of the background pulse. Data were not available to compare directly the two techniques. The device proposed here was tested using conventional means as the standard.

References

1. Arneson HN. Blood pressure amplifier with zero balancing means. US Patent No. 3, 868, 679, 1975.
2. Birnbaum MR. Blood pressure monitoring system. US Patient No. 3, 893, 452, 1975.
3. Collins NJ, Magora F. Sphygmomanometer: The indirect measurement of blood pressure. Anesth & Analg 1963; 42: 442–52.
4. Fiegel LJ. Blood pressure measuring. IBM Tech Disclosure Bull 1965; 8: 871–2.
5. Geddes LA: The direct and indirect measurement of blood pressure. Chicago, Illinois: Year Book Medical Publ 1970; 170–93.
6. Hill L, Flack M. On the method of measuring the systolic pressure in man and the accuracy of the method. Br Med J 1941; 1: 279–85.
7. McGouch AR, McDonald RD. Systolic-diastolic heart pressure monitor. Electron Eng 1981; 40: 673–7.
8. Mitchell RH, Ruff SC, Murnaghan GA. A system for determining long-term variations of blood pressure and pulse rate. IEEE Trans, BME-26: 310–311, 1979.
9. Mondshine W, Badger AS, Geddes LA. A numerical display device for the continuous indication of systolic and diastolic pressure. Med & Biol Eng 1969; 7: 95–7.
10. Ramsey M. Noninvasive automatic determination of mean arterial pressure. Med & Biol Eng & Comput 1979; 17: 11–18.
11. Sandman AM, Hill DW. An anlogue preprocessor for the analysis of blood pressure. Med & Biol Eng 1974; 12: 360–3.
12. Schulze AE, Meharg LS, Vogt FB. A system for the automatic measurement and digital display of systolic and diastolic blood pressure. Southwestern IEEE Conference Records 17F1–17F5, 1968.
13. Vachtsevanos GJ, Kalaitzakis KC, Papamarkos NJ, Ziakas GK, Economou KN, Gemitzis KC. Correlation study of arterial blood pressure level to the amplitude of the pressure pulse waveform. J Biomed Eng 1984; 6: 33–39.
14. Vachtsevanos GJ, Kalaitzakis KC, Papamarkos NJ. Design and development of a new electronic sphygmomanometer. J Med & Bio Eng & Comput 1985; 453–8.

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