PAPER

Automatic Correction of Left-Ventricular Pressure Waveform Using the Natural Observation Method

Jun-ichi HORI†, **Yoshiaki SAITOH**†, **Tohru KIRYU**†, *Members*
and **Taizo IIJIMA**††, *Honorary Member*

SUMMARY The pressure waveforms indicated on a catheter manometer system are subject to serious distortion due to the resonance of the catheter itself, or the compliance of a particular transducer. Although several methods have been proposed for improving those characteristics, they have never been put into practice. We have focused on the transfer function of the catheter manometer, and made a pilot system, using the natural observation method. This method has been suggested as a means of studying the structure of the instantaneous waveform. In this manner, we were able to increase the bandwidth in the frequency domain and reduce the ringing in the time domain. Correction was performed automatically, using a step wave. Reproduction of the waveform with a flushing device, was a task of equal simplicity, that allowed us to estimate the system parameters so that the response waveform became step-like. In the experiment, our system provided distortion-free left-ventricular pressure waveform measurements and exact evaluation of the cardiac pumping system. The values obtained came much closer to the original figures arrived at by the catheter-tip manometer system.

key words: *automatic correction; natural observation method; catheter manometer system; left-ventricular pressure waveform; flushing device*

1. Introduction

Both in monitoring and diagnosing problems in the cardiovascular system at the intensive or coronary care unit, it is necessary to measure the blood-pressure waveforms as accurately as possible. Measuring the blood-pressure waveforms with a catheter-tip manometer system (CTM) has commanded a lot of attention. The CTM yields more nondistorted signals than do conventional monitoring system. However, a distinct disadvantage of the CTM is that it is subject to immerse in blood clots, thus rendering it useless for longterm monitoring. On the other hand, the conventional catheter manometer system (CMS), which is generally used in clinical studies, involved serious frequency distortion due to the resonance of the catheter tube and the compliance of the transducer dome. Several methods for improving those characteristics have been realized, such as the Fourier analysis(1), lowpass filters(2)-(4) and the digital signal processing(5). Nevertheless, they have never been put into practice, because the frequency bandwidth was insufficient to represent the blood-pressure waveform, or because it was difficult to correct the waveform in real-time.

We have focused on the transfer function of the CMS and designed a means to compensate for errors, using the natural observation system (NOS)(6). The NOS has been proposed as a unique approach to the problem of reconstructing a signal without its Fourier series. We have already reported on the efficacy of the NOS, and the degree to which it improves the frequency response in an ambulatory electrocardiogram system(7). It is composed of a series of first-order high-pass filters, which are able to reconstruct an original waveform by the linear combination of outputs at each stage of filters.

In clinical studies engineering techniques should be performed simply in real-time. The NOS realized the inverse filter, which was limited in the higher-frequency band, with minimum adjusting parameters. That is to say, we improved the frequency characteristics of the CMS in the condition of reducing the high-frequency noise. We realized the automatic correction with digital filters in real-time. As a result, the noise-free and high-fidelity waveforms were in good agreement with those measured by the CTM.

After sampling signals through the analog-to-digital converter, we estimate the NOS parameters by the least square method. To execute the automatic correction, a step wave is available as a control signal. A step signal is easily produced by the flushing device. When the flushed waveform was detected by the appropriate threshold level, an original flushed waveform was subtracted from the blood-pressure waveform automatically. We then estimated the NOS parameters, so that the flushed waveform became a step waveform. Consequently, our system automatically adjusts its parameters, even during blood-pressure monitoring. Through simulated experiments, we confirmed the automatic correction in real-time, so that it compared with the distortion-free waveform obtained by the CTM. We also compared the evaluation indices in a heart pumping function before and after correction.

Manuscript received January 8, 1992.
Manuscript revised May 8, 1992.

† The authors are with the Faculty of Engineering, Niigata University, Niigata-shi, 950-21 Japan.
†† The author is with the Japan Advanced Institute of Science and Technology, Hokuriku, Kanazawa-shi, 920 Japan.

2. Method

2. 1 Catheter Manometer System

The transfer function of the CMS, an underdamped second-order dynamic system(8), can be described as follows:

$$H_c(s)=\frac{1}{s^2/\omega_n^2+2\zeta s/\omega_n+1}, \tag{1}$$

where s is the Laplace operator, ω_n is the natural angular frequency, and ζ is the damping factor. In a conventional CMS, output waveforms are distorted due to the low natural angular frequency and the small damping factor.

2. 2 Natural Observation Method

Figure 1 shows a block diagram of our method based on the NOS(9)-(11). The transfer function of the Mth-order NOS is given by

$$\begin{aligned} H_n(s)&=\sum_{i=0}^{M} a_i\left(\frac{s}{s+\omega_c}\right)^i \\ &=\left(\frac{\omega_c}{s+\omega_c}\right)^M \sum_{i=0}^{M}\frac{a_i s^i (s+\omega_c)^{M-i}}{\omega_c^M}, \end{aligned} \tag{2}$$

where ω_c is the cutoff angular frequency, and a_0, $a_1, \cdots$, a_M are the weighting parameters. We adjusted weighting parameters so that the $\sum$ term in Eq. (2) is identical to the denominator of Eq. (1). The transfer function of the total system, $H_t(s)$, is given by

$$H_t(s)=H_c(s)\cdot H_n(s)=\left(\frac{\omega_c}{s+\omega_c}\right)^M. \tag{3}$$

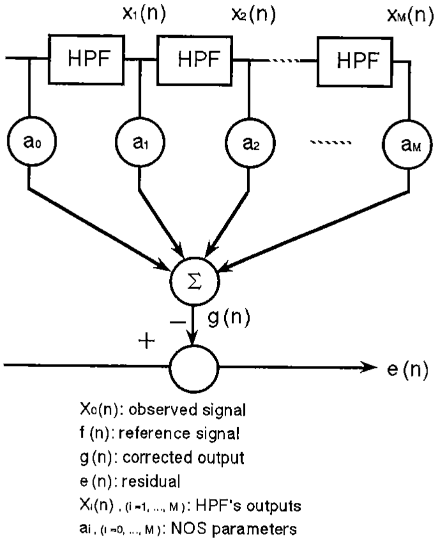


Fig. 1 Mth-order natural observation system for the correction of waveform.

Equation (3) consequently forms a Mth-order low-pass filter with the cutoff angular frequency ω_c. To improve frequency responses ω_c should be higher than the highest frequency component of blood-pressure waveforms: this procedure corresponds to the replacement of ω_n with ω_c.

Correction of the conventional CMS requires the second-order system because the transfer function of the CMS was observed as the second-order dynamic system indicated in Eq. (1). When we added an extension tube to the CMS, the order of the total transfer function might become more than two. In that case, the higher order of the correcting system, M, is required.

2. 3 Realization in Digital Filter

We acquired digital signals through the A/D conversion to estimate the NOS parameters. Each first-order high-pass filter was designed by the bilinear z-transformation. The corrected output, $g(n)$, described in Fig. 1 is given by

$$g(n)=\sum_{i=0}^{M} a_i x_i(n), \quad n=1, 2, \cdots, N \tag{4}$$

where $x_i(n)$, $i=1, 2, \cdots, M$, are output signals of individual high-pass filters and $x_0(n)$ is an input signal. N is the processing interval. Comparing Eq. (1) with Eq. (2), a_0 corresponds to 1. Assuming that $f(n)$ is a noise-free original signal, the residual, $e(n)$, is given by

$$\begin{aligned} e(n)&=f(n)-g(n) \\ &=f(n)-x_0(n)-\sum_{i=1}^{M} a_i x_i(n),\ n=1, 2, \cdots, N \end{aligned} \tag{5}$$

N covered the enough time for the transient part of the observed signal.

When $H_c(s)$ was noise-free system, NOS parameter vector $\boldsymbol{a}=[a_1, a_2, \cdots, a_M]^T$ could be calculated theoretically. But, the $H_c(s)$ involved noises in practice. The least square estimation determines the optimum NOS parameter vector $\boldsymbol{a}^*=[a_1^*, a_2^*, \cdots, a_M^*]^T$ by be following method:
Defining

$$\boldsymbol{f}=[f(1), f(2), \cdots, f(N)]^T,$$

$$\boldsymbol{x}_i=[x_i(1), x_i(2), \cdots, x_i(N)]^T,$$

$$\boldsymbol{e}=[e(1), e(2), \cdots, e(N)]^T$$

and

$$\boldsymbol{X}=[\boldsymbol{x}_1, \boldsymbol{x}_2, \cdots, \boldsymbol{x}_M],$$

then the Eq. (5) is rewritten as

$$\boldsymbol{e}=\boldsymbol{f}-\boldsymbol{x}_0-\boldsymbol{X}\boldsymbol{a}. \tag{6}$$

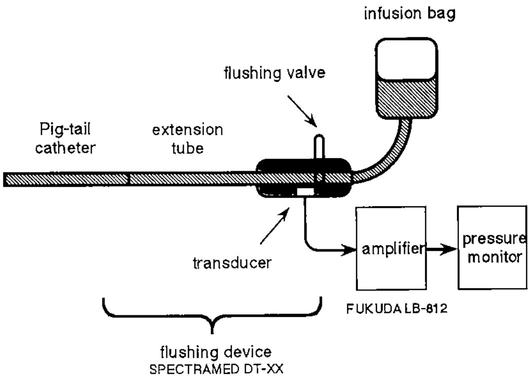


Fig. 2 Catheter manometer system with the flushing device. The flushing operation produced a step-like pressure waveform varying from negative to zero.

The evaluation function J is given by

$$J=(\boldsymbol{f}-\boldsymbol{x}_0-\boldsymbol{X}\boldsymbol{a})^T(\boldsymbol{f}-\boldsymbol{x}_0-\boldsymbol{X}\boldsymbol{a}). \qquad (7)$$

By $\partial J/\partial \boldsymbol{a}=\boldsymbol{0}$, the normal equation is

$$\boldsymbol{X}^T\boldsymbol{X}\boldsymbol{a}=\boldsymbol{X}^T(\boldsymbol{f}-\boldsymbol{x}_0). \qquad (8)$$

The optimum NOS parameter vector, $\boldsymbol{a}^*$, is calculated using the following equation.

$$\boldsymbol{a}^*=(\boldsymbol{X}^T\boldsymbol{X})^{-1}\boldsymbol{X}^T(\boldsymbol{f}-\boldsymbol{x}_0). \qquad (9)$$

The correction was achieved with the minimum number of parameters using the NOS.

2. 4 Flushing Device

To estimate the NOS parameters, we require an observed signal from the CMS and a noise-free signal that has to be known in advance. We used a step wave as a control signal. A step wave is easy to produce with a flushing device(12), which is usually used to remove bubbles from the fluid-filled catheter. Figure 2 shows the CMS with the flushing device attached. The flushing device consists of a flushing valve and a pressure transducer. It connects to a pressurized infusion bag that is filled with saline solution. The flushing operation originally produced a step-like pressure waveform varying from negative to zero. Due to the poor frequency characteristics, the flushed response usually had ringing after flushing. We estimated the system parameters so that the flushed waveform became similar to the original step-like one.

2. 5 Parameter Estimation During Blood Pressure Measurement

When we operate the flushing device during blood-pressure measurement, the flushed waveform is superimposed on the blood-pressure waveform. Figure 3 shows the parameter estimation procedure used during measurement. When a step response was detected by an appropriate threshold level, a step response waveform with additive blood-pressure waveform, A, was separated from the blood-pressure waveform, B, by segmentation. We also adjusted the timing for phases of both waveforms. When B is subtracted from A, A-B indicates the desired waveform. The components except the flushed waveform in A should be equal to the components in B. That is to say, the blood-pressure waveforms must be periodic. We confirmed that fluctuations of rhythm in blood-pressure had no influence on our procedure.

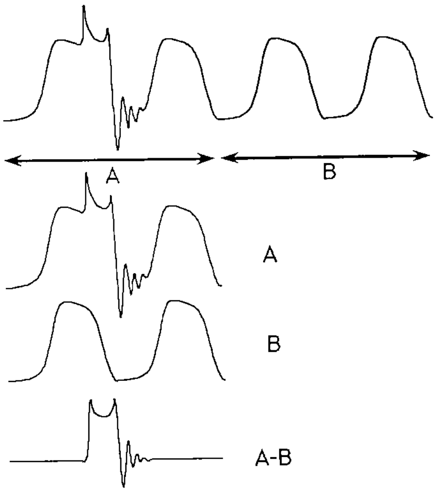


Fig. 3 Parameter estimation during blood-pressure measurement. A step response with additive blood-pressure waveform, A, was separated from the blood-pressure waveform, B. A-B indicated the desired waveform.

3. Experimental Results

3. 1 Results Corrected in Real Time

Figure 4 shows a block diagram of the automatic correcting system consisting of digital filters. We digitized pressure signals with a 12 bit analog-to-digital conversion at a sampling rate of 500 Hz. When a step waveform is detected, the procedure of waveform segmentation and parameter estimation, the upper sequence block of Fig. 4, is automatically performed. Then the parameters are updated in the correcting system of the lower sequence block. It took about 3s from flushing operation to updating the correcting system. After that, the correction was executed in real-time.

Figure 5 shows the experimental procedure using a pressure simulator (BIO-TEK Model 601A) and a flushing device (SPECTRAMED DT-XX). High fidelity left-ventricular waveforms detected with a Miller CTM were fed into the pressure simulator. The pressure simulator transformed an electrical signal into a pressure signal. We measured the simulated pressure signals with a Pig-tail CMS and corrected the distorted

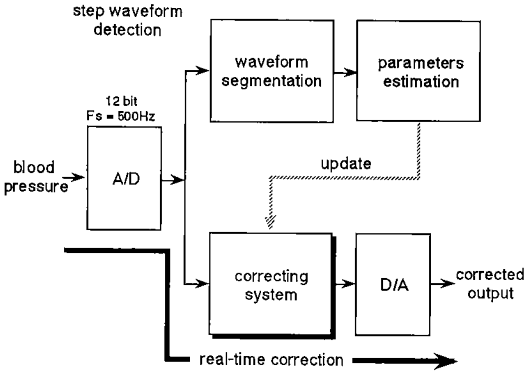


Fig. 4 Automatic correcting system. When a step waveform is detected, the upper sequence block is automatically performed. After updating the correcting system, the correction was performed in real-time.

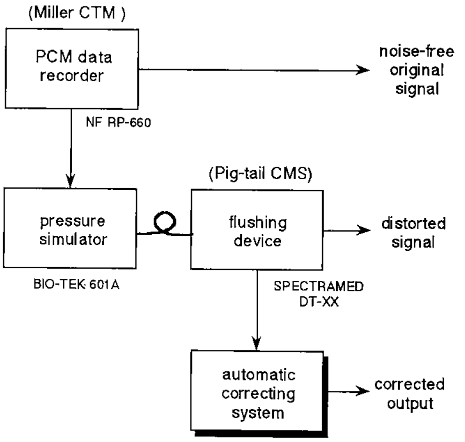
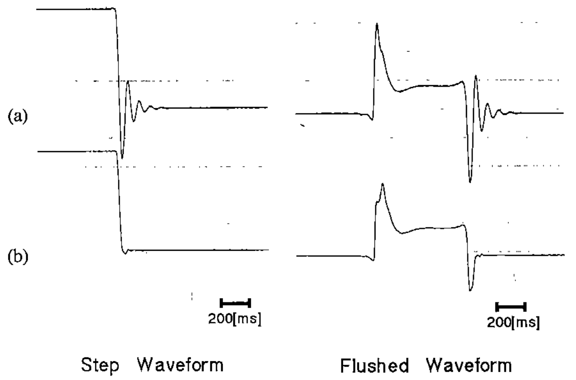


Fig. 5 Experimental procedure using a pressure simulator and a flushing device. The pressure simulator transformed an electrical signal into a pressure signal. We measured the simulated signals with a Pig-tail CMS and corrected the distorted pressure waveforms.

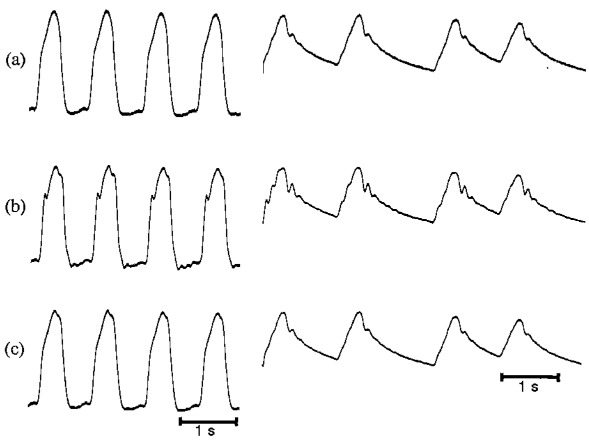
pressure waveforms.

Figure 6 shows step waveforms and flushed waveforms, before and after correction. The cut-off frequency ($=\omega_c/2\pi$), and the order, M, of the correcting system is equal to 50 Hz and 6th, respectively. The processing interval, N, is equal to 300 points (the processing time corresponds to 0.6 s). The step response of the Pig-tail CMS (Fig. 6(a)) was followed by ringing. After correcting the step response (Fig. 6 (b)), the ringing was suppressed to the point where it became virtually flat. This resulted from the compensation of frequency characteristics by Eq. (3). The ringing was observed in the flushed response as in the step response. Applying the same $H_n(s)$ to the flushing response observed by the Pig-tail CMS, the ringing after flushing was improved to be flat.



Step Waveform Flushed Waveform

Fig. 6 Corrections of a step waveform and a flushed waveform. (a) Response waveforms of the CMS. (b) Corrected outputs of (a). Both response waveforms were corrected with same NOS parameters.



Left-ventricular pressure waveforms in normal condition.

Aortic pressure waveforms in arrythmia.

Fig. 7 Corrected results of pressure waveforms in real-time. (a) Original pressure waveforms measured with a Miller CTM. (b) Distorted waveforms measured with a Pig-tail CMS. (c) Corrected outputs of (b).

3.2 Correction and evaluation of the Blood-pressure Waveforms

Figure 7 shows the results of correction in clinical studies. The left panel shows left-ventricular pressure waveforms in normal rhythm and the right panel shows aortic pressure waveforms in arrhythmia. Every kind of waveforms were corrected by the NOS. Since corrected waveforms showed no time-lag, we were able to correct the pressure waveforms with a Pig-tail CMS, in real-time.

Figure 8 shows left-ventricular pressure waveforms in a human heart and the evaluation indices. The pressure waveform and the rate of pressure rise (*dP*/*dt*) in the Pig-tail CMS (Fig. 8(b)) have lightly damped oscillations, compared with those illustrated in Fig. 8(a), as obtained by the Miller CTM. Correct-

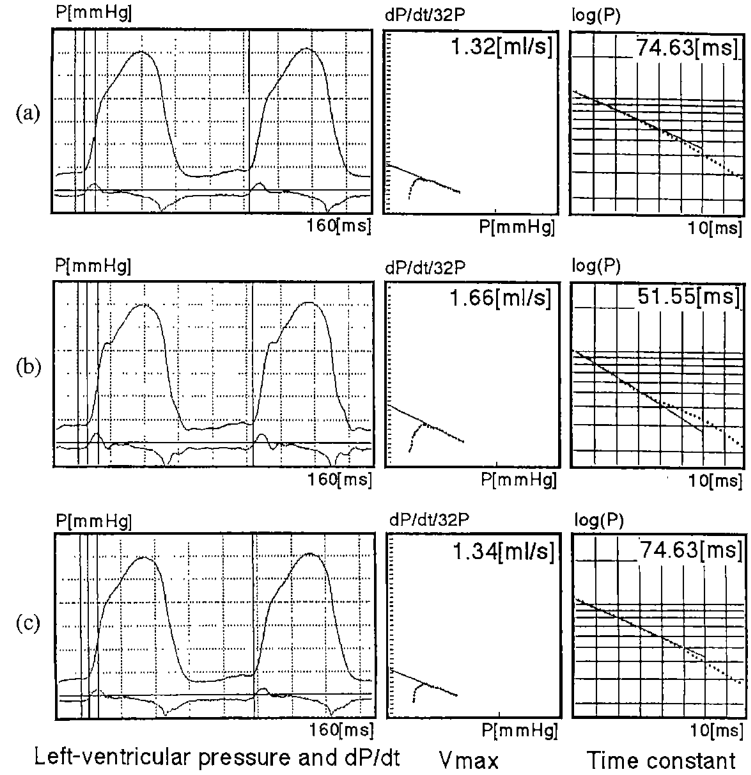


Fig. 8 Correction and evaluation of left-ventricular pressure waveforms. (a) Original pressure waveforms measured with a Miller CTM. (b) Distorted waveforms measured with a Pig-tail CMS. (c) Corrected outputs of (b). The maximum velocity of the contractile element, $V_{\max}$, and the time constant indicate the systolic and diastolic function of an intact human heart, respectively.

ed outputs (Fig. 8(c)) show remarkable improvement, and they were in good agreement with the waveforms described by the Miller CTM.

Evaluation indices calculated are shown in the center and right panel of Fig. 8. The maximum velocity of the contractile element ($V_{\max}$) is an index of the contractile state of an intact human heart. It was calculated from the pressure-velocity curve during a isovolmic systole(13). The time constant (TC) is an index of the isovolmic diastole function after ejection. It was calculated by fitting an exponential function to the pressure-fall from the time instant of the maximum negative dP/dt(14). $V_{\max}$ and TC could not be calculated as accurately with the pressure waveform in the Pig-tail CMS, as they could by the Miller CTM. The estimates by our method were similar to those arrived at by the Miller CTM.

4. Discussion

Overshoot minimization techniques for filters were proposed by several researchers. Kuo reduced this figure by adding a constant, times the second derivative of the response, to the original response(15). Reck and Melvin investigated an optimum value for the constant used in the above technique(16). Suhash and Dutta replaced the second derivative with a series of first and second derivatives(17). A defect of these filters, however, was their susceptibility to the interference of a random noise.

The band-limiting technique, realized by a biquad filter(18) or an autoregressive moving average model also decreased the unnecessary high-frequency noise. The transfer functions of the above filters are constructed in the same manner as the transfer function shown in Eq. (2). Implementation of the NOS is different from these other methods. The NOS is a simple composition of successive first-order high-pass filters, where the number of parameters adjusted is limited to the minimum number in the filter of Eq. (2). In clinical studies implementation of engineering techniques should be as simple as possible. The band-limited inverse filter with the minimum number of parameters was implemented only using the NOS. Thus, the NOS proposed in this paper is, we believe, most appropriate and applicable for the situation.

Our pressure measuring system, or the CMS, consists of a flushing valve and a disposable pressure transducer. If we depress the flushing valve button, the entire catheter tube and transducer dome are pressurized to a high level. When the button is released, the negative pressure is reversely applied to the tip of the catheter, thereby creating a step-like waveform that varies from negative to zero. Since the system function of a linear filter is represented by the step response, the system function of the CMS can be estimated by the step-like waveform produced by the flushing operation.

During long term blood-pressure monitoring, the condition of the CMS may vary. For example, when air bubbles are generated in a catheter, the frequency characteristics tend to deteriorate. In such cases, the correcting system should be updated, in order to maintain the optimum condition or weighting parameters, even during pressure measurement. We accomplished this only pushing flushing: the update of weighting parameters follow the description of the flushing procedure. We also confirmed that fluctuations of rhythm in blood-pressure had no influence on our correction technique.

To evaluate a cardiac pumping function, it is necessary to measure the left-ventricular pressure waveform, with as high fidelity as possible. Recently, the catheter examination continued through few days in clinical study. Thus, the pressure measurement requires not only high-fidelity but also a high anticoagulant factor. High anticoagulant factor refers to the tip of the catheter becoming only slightly immersed in blood clots. Since a measuring system of the pressure waveform satisfying both conditions is as yet unavailable, either the anticoagulant of the CTM should be increased, or the frequency characteristics of the conventional CMS should be improved. In the former case, the CTM would be subject to coagulate blood clots during long term monitoring because of its complicated structure. Alternatively, we improved the frequency characteristics of the conventional CMS and achieved distortion-free measurement of the left-ventricular waveform.

$V_{\max}$ and TC are typical evaluation indices calculated from the left-ventricle pressure waveform. By our technique, we were able to obtain values of much greater accuracy from a conventional CMS, an achievement that will, in the very near future, enable us to easily monitor patients in intensive, and coronary care unit.

5. Conclusion

Using the NOS, we corrected the distorted blood-pressure waveforms obtained by conventional CMS. Digital filters were incorporated in the NOS, and the correction was accomplished in real-time. Automatic correction is performed by the least square estimation and the automatic segmentation of the flushed response. Our system provides distortion-free blood-pressure waveform and, therefore, exact evaluation of the cardiac pumping system. With an anticoagulant CMS, we shall be able to obtain high-fidelity waveforms, even during long-term monitoring.

Acknowledgement

The authors wish to thank Drs. T. Matsukawa and R. Hashimoto and Mr. Y. Hasegawa and K. Itoh for their experimental help and advice.

References

(1) Nimura Y., Tachibana T., Ichie Y., Takeuchi S., Yasui S., Mizuno Y. and Hisada S.: "Analysis of Left Ventricular Waveform and Characteristic of Catheter-Manometer System", Jap. J. Med. Elec. & Biol. Eng., **13**, 4, pp. 214–222 (1975).

(2) Melbin J. and Spohr M.: "Evaluation and Correction of Manometer Systems with Two Degrees of Freedom", J. Appl. Physiol., **27**, 5, pp. 749–755 (1969).

(3) Gould K. L., Trenholm S. and Kennedy J. W.: "In Vivo Comparison of Catheter Manometer Systems with the Catheter-Tip Manometer", J. Appl. Physiol., **34**, pp. 263–267 (1973).

(4) Falsetti H. L., Mates R. E. and Carroll R. J.: "Analysis and Correction of Pressure Wave Distortion in Fluid-Filled Catheter Systems", Circulation, **50**, pp. 165–172 (1974).

(5) Takeuchi H., Toraichi K., Kamada M. and Mori R.: "Analysis of Left Ventricular Pressure by B-spline Windows", Proc. Jap. J. Med. Elec. & Biol. Eng., **24**, p. 454 (1986).

(6) Saitoh Y., Hori J., Kiryu T., Makino H., Tamura K., Yoshizaki T. and Iijima T.: "Correction of Pressure Waveforms in a Catheter Manometer System by Natural Observation Method", Jap. J. Med. Elec. & Biol. Eng., **26**, 3, pp. 133–138 (1988).

(7) Hori J., Saitoh Y., Kiryu T., Asakawa T., Tamura K. and Iijima T.: "Improvement of the Frequency Responses in an Ambulatory ECG System by the Natural Observation Method", Frontiers Med. Biol. Eng., **2**, 2, pp. 134–145 (1990).

(8) Fry D. L.: "Physiologic Recording by Modern Instruments with Particular Reference to Pressure Recording", Physiol Rev., **40**, pp. 753–788 (1960).

(9) Iijima T.: "Fundamental Theory on Natural Observation of Wave Form", Trans. IEICE **J67-A**, 10, pp. 951–958 (Oct. 1984).

(10) Iijima T.: "Theoretical Study on Reconstruction of Wave Forms by a Natural Observation Filter", Trans. IEICE, **J74-A**, 3, pp. 430–434 (March 1991).

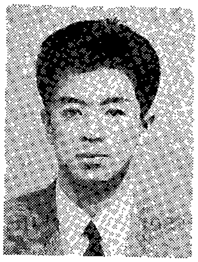
(11) Iijima T.: "A Natural Observation Filter Acceptable for Almost Periodic Wave Forms", Trans. IEICE, **J74-A**, 3, pp. 435–441 (March 1991).

(12) Gardner R. M., Warner H. R., Toronto A. F. and Gaisford W. D.: "Catheter-Flush System for Continuous Monitoring of Central Arterial Pulse Waveform", J. Appl. Physiol., **29**, 6, pp. 911–913 (1970).

(13) Mason D. T., Spann J. F. and Zelis R.: "Quantification of the Contractile State of the Intact Human Heart", Am. J. Cardiol., **26**, pp. 248–257 (1970).

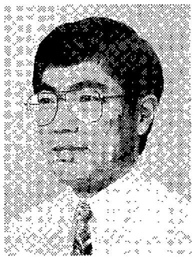
(14) Weiss J. L., Frederiksin J. W. and Weisfeldt M. L.: "Hemodynamic Determinants of the Time-Course of Fall in Canine Left Ventricular Pressure", J. clinical investiga-

tion, **58**, pp. 751-760 (1976).
(15) Suhash C. and Dutta R.: "Reducing Overshoot and Ringing in Filters", Proc. IEEE, pp. 1189-1190 (1966).
(16) Reck R. H. and Melvin D. W.: "An Overshoot Minimization Technique for Low-Pass Filters", IEEE Trans. Circuit Theory, **CT-20**, 2, pp. 158-161 (March 1973).
(17) Kuo F. F.: "A Method to Reduce Overshoot in Filters", IRE Trans. Circuit Theory, CT-9, pp. 413-414 (Dec. 1962).
(18) Thomas L. C.: "The Biquad. Part 1—Some Practical Design Considerations", IEEE Trans. Circuit Theory, **CT-18**, 3, pp. 350-357 (May 1971).



Jun-ichi Hori was born in Yamagata Prefecture, Japan, in 1963. He received the B.E. and M.E. degrees in information engineering from Niigata University, Niigata, Japan, in 1986 and 1988, respectively. In 1988 he joined the staff of the Department of Information Engineering, Niigata University, as an Assistant. His research interests include the improvement of biomedical instrumentation and the analysis of biomedical signals. Mr. Hori is a member of the Japan Society of Medical Electronics and Biological Engineering and The Japanese Society of Artificial Organs.

Yoshiaki Saitoh was born in Niigata Prefecture, Japan. He received the B.E. degree in electrical engineering from Niigata University, Niigata, Japan, in 1963, and the M.E. and Ph.D. degrees in electrical engineering from Hokkaido University, Sapporo, Japan, in 1965 and 1970, respectively. In 1965 he joined the staff of the Department of Electronics, Niigata University, as an Instructor. Since 1980, he has been a Professor in the Department of Information Engineering at Niigata University. His recent research interests include measurements and stimulation of the human organs, hyperthermia systems, and biomedical signal processing. Dr. Saitoh is a member of the Japan Society of Medical Electronics and Biological Engineering.



Tohru Kiryu was born in Niigata Prefecture, Japan, in 1952. He received the B.E. and M.E. degrees in electronics engineering from Niigata University, Niigata, Japan, in 1975 and 1977, respectively and the Dr. Eng. degree in computer science from Tokyo Institute of Technology, Tokyo, Japan, in 1985. He joined the school of dentistry of Niigata University, from 1977 to 1978 as an Assistant. Since 1978 he has been with the Department of Information Engineering, Niigata University as an Assistant, and since 1986 as an Associate Professor. The Department of Information Engineering corresponds to the computer science. He has been working on biomedical signal processing, especially time-varying spectral analysis, time-varying parameters estimation using nonstationary models and nonstationary stochastic process. For applications, his interests have been in myoelectric signal analysis during dynamic movements and speech signal modeling around transient parts. Dr. Kiryu is a member of the Japan Society of Medical Electronics and Biological Engineering and the Japan Prosthodontic Society.

Taizo Iijima was born in Osaka Prefecture, Japan, in 1925. He graduated in 1948 from Dept. of Electrical Engineering, Tokyo Institute of Technology, and joined the Electrotechnical Laboratory the same year. Since then, he has been engaged in research on electromagnetic field theory, pattern recognition theory, image processing and speech recognition, as well as development of optical character reader (OCR). He also received Dr. Eng. degree in 1957 from Tokyo Institute of Technology. In 1972 he became a Professor at Tokyo Institute of Technology, and in 1986, Professor Emeritus. He then became a Professor at Tokyo Engineering University, and in 1991, Professor Emeritus. From 1991 he has been a Vice President at Japan Advanced Institute of Science and Technology, Hokuriku. Dr. Iijima is a member of the Information Processing Society of Japan, and Acoustical Society of Japan. He received an Achievement Award in 1977 and Paper Awards in 1965, 1973, 1976, 1982, respectively, and in 1989 an Accomplishment Award, from the Institute of Electronics, Information and Communication Engineers of Japan. Furthermore, he received several awards from the Ministry of National Trade and Industry. In 1990, he received a Medal of Honor with Purple Ribbon, from Emperor.