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Estimating Input Original Waveform from Catheter-Manometer System Output Waveform

− An Experimental Study with Pulmonary Artery Catheter −

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The present study was performed to investigate the precision of and problems in arithmetic operations for synthesis/estimation of the input waveform from the output waveform of a pulmonary artery catheter (PAC) manometer system, by imposing arterial pressure waveform as an input pressure waveform. When a PAC manometer system is regarded as a second-order system, the input waveform can be synthesized/estimated from the output waveform by determining the natural frequency (fn), and damping coefficient (ζ) of the said system. The precision of synthesis/estimation diminished with the decreasing fn of the PAC manometer system. An attempt was thus made to explore the cause of the reduction in precision of estimation due to decreases in fn by constructing a second-order system equivalent circuit and a differentiation arithmetic circuit on the circuit simulator. The results demonstrated that correct synthesis/estimation of input waveform were attainable only when the set of fn and ζ for the arithmetic circuit coincided with that of the PAC manometer system. Decreases in fn correspond to such instances where the PAC manometer system can no longer be regarded as a second-order system, thus nullifying the measured fn and ζ. This accounts for the reduction in precision.

Key words: arterial pressure wave, catheter manometer system, second-order system, frequency characteristics, arterial waveform reconstruction

INTRODUCTION

The catheter-manometer system displays responses depending on the frequency of input signals (blood pressure waveform). This, termed frequency characteristics of the system, accounts primarily for distortion/deformation of output waveform. We have reported that the degree of distortion/deformation of output waveform is quantitatively assessable in terms of the two parameters of the catheter-manometer system, natural frequency and damping coefficient [1, 4]. The key to achieving waveform fidelity consists in assemblage of a catheter-manometer system so as to maintain natural frequency as high as practicable and to minimize damping coefficient. However, various limitations are inherent in the practice of manometric measurement in clinical settings. Waves distorted on amplitude axis may lead to errors in pressure readings and waves distorted on time axis may cause undue delay when compared with other parameters such as ECG or arterial pressure. Furthermore, accuracy of depicting transient phenomena is of great importance as in pulmonary artery occlusion to estimate pulmonary capillary pressure. If it is feasible to estimate true input waveforms from output waveforms under such circumstances, the purpose of leading off pressure waves may be accomplished irrespective of waveform evaluation. Frequency characteristics of the catheter-manometer system, as mentioned above, can be expressed in terms of natural frequency and damping coefficient. It is easy to measure these two parameters in the laboratory setting as well as in clinics, and this report describes a method for estimating the input waveform from the output waveform obtained from the catheter-manometer system and the precision of the estimation.

THEORETICAL BACKGROUND FOR THE ESTIMATION METHOD

The catheter-manometer system is expressed in terms of LRC circuit comprising mass (L, inductance), viscosity (R, resistance) and elasticity (C, compliance) as illustrated in Figure 1.
The input, y, can be expressed as:

$$y = L \cdot di/dt + Ri + (1/C) \int i dt$$

Wherein i represent the circuit current. When the voltage between both ends of compliance C is given as output e, then e represents the output of the manometer system and the following equation holds [6].

$$y = LC \cdot d^2e/dt^2 + RC \cdot de/dt + e \cdots\cdots\cdots\cdots\cdots\cdots (1)$$

By Laplace transformation of this equation, the transfer function, G(s), is given as

$$G(s) = e(s)/y(s) = 1/(LCs^2 + RCs + 1) \cdots\cdots\cdots\cdots (2)$$

When the natural frequency and damping coefficient of the secondary series are expressed as ω_n and ζ, respectively, the general formula can be given as

$$G(s) = 1/\{(1/\omega_n^2)s^2 + (2\zeta/\omega_n)s + 1\} \cdots\cdots\cdots\cdots\cdots (3)$$

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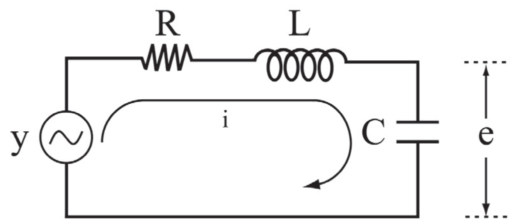

Fig. 1 Catheter-manometer system model. The catheter-manometer system can be expressed as an LRC circuit comprising inertia, viscosity and elasticity of liquid mass, connected in series, within the said manometer system.

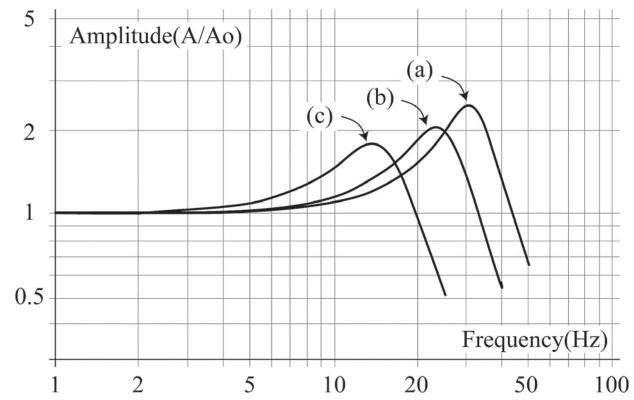

Fig. 2 Three types of frequency characteristics of the catheter-manometer system. Filling conditions of catheter: (a) filled with deaerated water, (b) filled with tap water at ordinary temperature (26°C), or filled with tap water at ordinary temperature and then warmed (to 37°C). The fn and ζ calculated from Equation 5 were (a) 32.2 Hz and 0.21, (b) 25.0 Hz and 0.25, and (c) 15.2 Hz and 0.29, respectively.

The two coefficients LC and RC in the differential equation and transfer function eventually correspond to $1/\omega_n^2$ and $2\zeta/\omega_n$, respectively. Therefore, Equation 1 can be expressed as:

$$y = (1/\omega_n^2) \cdot d^2e/dt^2 + (2\zeta/\omega_n) \cdot de/dt + e \quad \cdots\cdots (4)$$

It is feasible to measure the ω_n and ζ with practical precision, nevertheless, from frequency characteristics in laboratory settings or by the flushing method or tapping method in clinical settings, so that the coefficients in Equation 4 can thereby be determined. The left member of Equation 4 represents the input waveform while the right member consists of a sum of the terms of output waveform and its first and second derivatives, thus indicating that it is possible to estimate the input waveform by differential arithmetic of the output waveform. In other words, if observed blood pressure waveform is differentiated twice by some means after determining ω_n (= $2\pi \cdot$fn) and ζ of the manometer system, the original intravascular pressure, being the input of the manometric system, can be computed by adding up those derivatives.

METHODS AND OBJECTS

(1) Measurement of fn and ζ of the manometer system

A standard pulmonary artery catheter (quadruple lumen, No. 7 Fr., 110 cm) was coupled to a pressure generator (601A, Bio Tec). After determining the frequency characteristics of the system in the 1-60 Hz band by imposing the sinusoidal wave vibration of the generator onto the catheter, fn and ζ were calculated from the two frequency characteristics, resonance frequency (fr) and resonance amplitude (Ar) using the following expressions [2].

$$\zeta^2 = (1 \pm (1\text{-}1/Ar^2)^{1/2})/2, \ fn = fr/(1\text{-}2\zeta^2)^{1/2} \quad \cdots\cdots (5)$$

Frequency characteristics were derived under three different filling conditions of pulmonary artery catheter, i.e., (a) filled with deaerated water, (b) filled with tap water at ordinary temperature (26°C), or filled with tap water at ordinary temperature and then warmed (to 37°C).

(2) Arithmetic operation of differentiation

Differential output is proportional to the frequency components of the subject waveform. A low-pass filter or moving average processing is essential for the output because an abundance of high-frequency noise is inherent in the output waveform. The smoothing differentiation technique based on polynominal fitting is employed in this study. With this technique, it is practicable to attain a smooth differential waveform to adjacent points as the procedure utilizes a differential term obtained by fitting a polynominal primarily encompassing a differential point, hence containing elements of the moving average. When the output waveform data series and the differential coefficient at that point are assigned x (i) and y (i), respectively, the actual arithmetic operation will be a convolution arithmetic operation of x (i) and Savitzky-Golay coefficient sequences, w (j), so we get the following expressions [4]:

$$y(i) = (1/W) \cdot \sum x(I + j) \cdot w(j), \ W = \sum w(j)$$

(3) Estimation of waveforms

Artery pressure waveforms recorded on a data recorder were fed into a pressure generator, and the waveforms thereby generated were then applied as input waveforms to a catheter. For each of the above-mentioned three types of frequency characteristics, the input waveform and the output waveform from the catheter were loaded into a computer at a resolution of 12 bits and a sampling frequency of 4 kHz. The output waveform loaded was subjected to the differential processing mentioned above, followed by waveform synthesis according to Equation 4 to obtain an estimated waveform of the input. The difference between the input waveform and the estimated waveform was computed to evaluate the precision of estimation.

RESULTS

The frequency characteristics derived under three different filling conditions of pulmonary artery catheter, i.e., (a) filled with deaerated water, (b) filled with tap water at ordinary temperature (26°C), or filled with

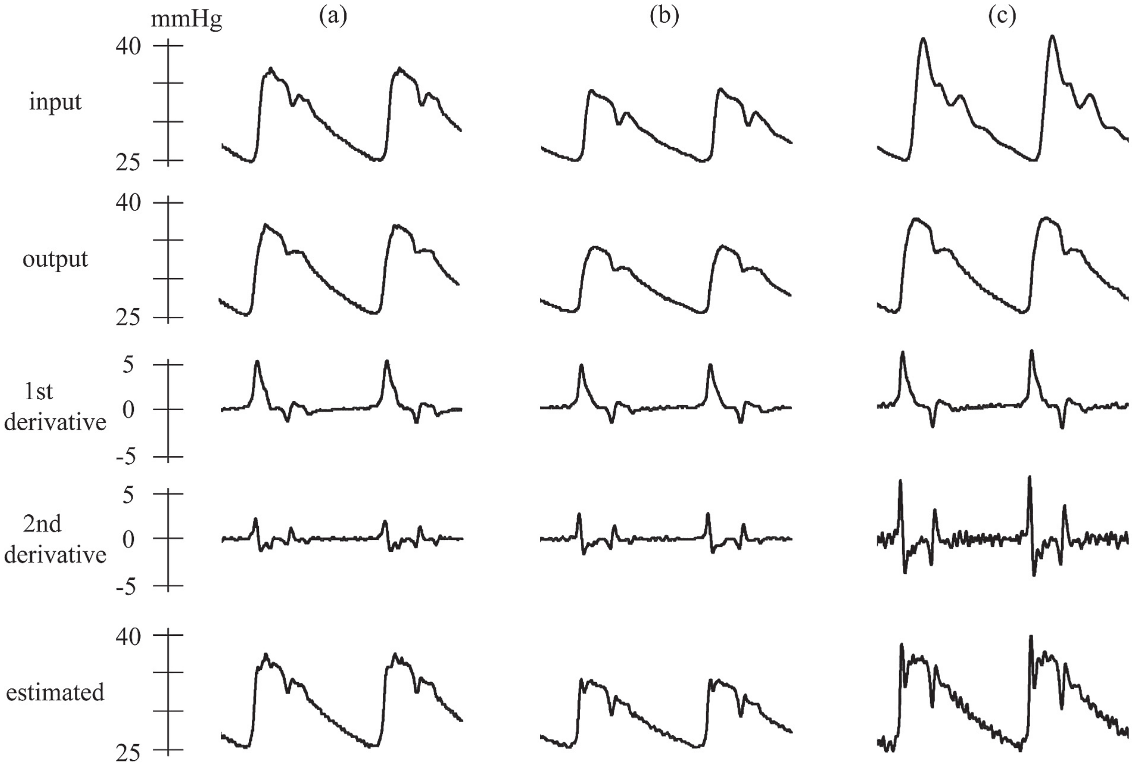

Fig. 3 Synthesis and estimation of input waveform. Letters (a), (b) and (c) correspond to the three types of frequency characteristics. From the top downward in each panel, the data represent the catheter input waveform (input), the catheter output waveform (output), the first differential derivative of output waveform (1st derivative), the second differential derivative of output waveform (2nd derivative), and the estimated waveform of input synthesized from the first and second derivatives (estimated).

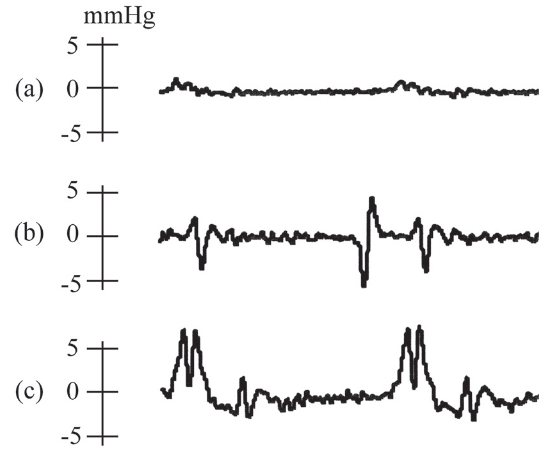

Fig. 4 Estimated errors. Letters (a), (b) and (c) correspond to the three types of frequency characteristics. Presented are waveforms of differences between the input waveform and the estimated waveform. As can be seen, the amplitude of difference waveform increased progressively in order of (a), (b) and (c), that is, with decreasing natural frequency.

tap water at ordinary temperature and then warmed (to 37°C), are shown in Figure 2. The fn and ζ calculated from Equation 5 were (a) 32.2 Hz and 0.21, (b) 25.0 Hz and 0.25, and (c) 15.2 Hz and 0.29, respectively.

Figure 3 illustrates the process of input synthesis/estimation from the catheter output waveform for the respective three types of frequency characteristics. From the top downward in the figure, the data represent the catheter input waveform (input), the catheter output waveform (output), the first differential derivative of output waveform (1st derivative), second differential derivative of output waveform (2nd derivative), and the synthetic estimated waveform of input (estimated). The estimated waveforms, though with overlapping fine oscillations due to the differentiation manipulation, reproduce fairly well the gross changes in input waveforms.

Figure 4 presents waveforms of differences between the input waveform and the estimated waveform under the above-mentioned conditions. As can be seen, the amplitude of difference waveform increased progressively with decreasing natural frequency, that is, the chart indicates that the difference between input and output increased and the precision of estimation diminished with decreasing natural frequency.

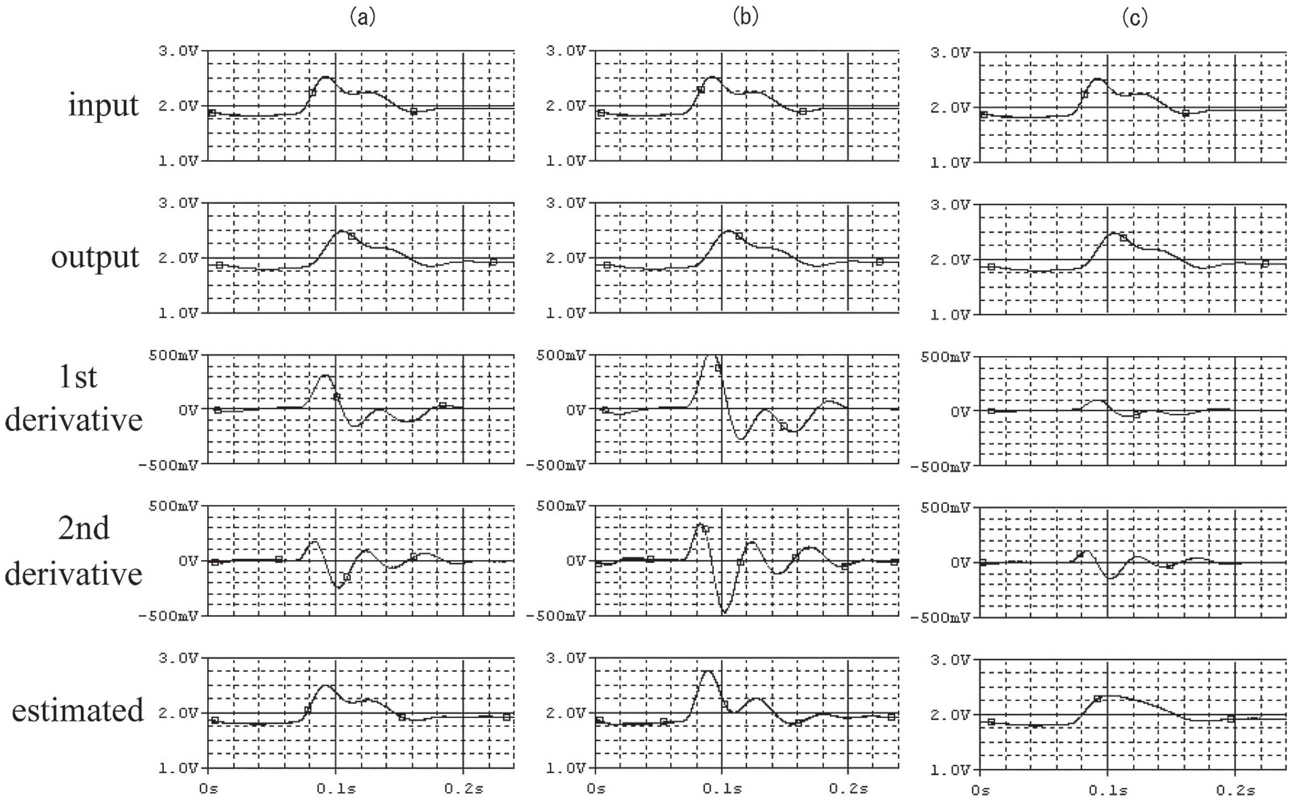

Fig. 5 Synthesis of input waveform by simulation. The fn and ζ values set for the secondary system circuit are equal to those set for the arithmetic circuit (a) or different from the latter [(b) and (c)]. Processes of waveform synthesis were identical with those in Figure 3. It is obvious that correct synthesis/estimation of input waveform cannot be achieved if fn and ζ values differ between the two circuits. See the text for details.

DISCUSSION

Attempts to estimate input waveform from an output waveform obtained with a measurement system upon determination of its transfer characteristics by some means have been reported with various methods for a variety of objects [3, 5]. For the primary delay system, as is the case with capnometer responses, the equation $y = RC \cdot de/dt + e$ holds to depict the relationship between input (y) and output (e). The coefficient RC for the first differential derivative term corresponds to the time constant T of a step response. Therefore, measurement of T permits correction of output response, namely, synthesis or estimation of input waveform [5]. Responses of the catheter-manometer system described herein are of a secondary system, so that the second differential derivative term $LC \cdot d^2e/dt^2$ is added to the primary delay system as shown in Equation 1, yet the same basic concept of input waveform estimation holds. As seen from Equation 4, it would be feasible to synthesize or estimate the input waveform insofar as the natural frequency and damping coefficient of the catheter-manometer system can be determined by some means irrespective of the values of L, C and R.

Such estimation of the input waveform requires differential operations for the output waveform. It is not always easy, however, to carry out the intended selective differentiation on a computer. Differential operations may cause prevalence of high-frequency noise in the output in proportion to frequency components of the subject waveform, thereby making it difficult to discern the intended differential waveform. Differentiation thus is an arithmetic operation preferably to be avoided, but we think that it is inevitable in the present input estimation where differential processing per se constitutes its essence. A smoothing differentiation with polynominal fitting applied is adopted in the present method. Since the method utilizes the differential coefficient by regarding a plurality of adjoining data series within the subject waveform as a polynominal, a smooth differential coefficient series to which a moving average element is essentially added is obtained. The polynominal fitting employs the least squares procedure and hence involves a vast amount of calculation. However, a simple convolution operation suffices as the table of Savitzky *et al.* is utilized as mentioned above. The blood pressure waveform data processing in this method may be adequately practicable as a real-time operation. With this method of processing, as illustrated in Figure 3, major pressure change components are extracted in the form of first and second differential waveforms while disregarding minor changes in catheter output waveform.

As seen from the results presented in Figures 3 and 4, the error in peak value of waveform (systolic pressure) was approximately 5% at the lowest fn of 15.2

Hz. As this fn value represents an average with the use of standard catheters, we may well infer that the true intravascular pressure can be estimated with an error not greater than a few percent unless the blood pressure waveform led off is extremely dulled or in a vibratory state due to resonance.

However, as evident from the results, the precision of waveform synthesis/estimation diminished progressively with decreasing fn of the catheter-manometer system. It is necessary to impose varied conditions on the catheter-manometer system in order to examine its cause, but it is difficult to set fn and ζ at arbitrary values for an actual catheter-manometer system. In view of this, we prepared an electrically equivalent circuit of a secondary system that enabled arbitrary fn and ζ setting, integrated it into an electrical circuit simulator (Pspice10, Cadence Design Systems) along with an arithmetic circuit comprising differentiation and addition, and examined factors that might have bearing on the synthesis and estimation of input waveform. An arterial pressure waveform data file was taken as a secondary system input waveform. Values of fn and ζ set for these secondary system circuits correspond to fn and ζ in the catheter-manometer system. For details of the circuit constitution, readers are referred to the literature [2, 8]. Results of the simulation are shown in Figure 5. The same values of fn and ζ were set for secondary system circuits (a), (b) and (c) (i.e., fn = 20 Hz, ζ = 0.55). Eventually, the output waveform (output) was equal in all the three circuits. The values of fn and ζ set for the arithmetic circuit in (a) were equal to those set for the secondary circuit, while the values of fn and ζ set for the arithmetic circuit in (b) and (c) were combinations that gave the same peak frequency (which means the highest frequency component that would yield an output waveform faithful to an input waveform within the permissible limits of error, and expresses the fidelity of the output waveform) (fn = 15 Hz, ζ = 0.68; and fn = 25.5 Hz, ζ = 0.24, respectively). As obvious from (b) and (c), any correct synthesis/estimation of input waveform cannot be achieved if fn and ζ values of the secondary system mutually vary from those of the arithmetic circuit. Correct synthesis of input waveform is hardly accomplished even if fn and ζ values are modified in an attempt to give the same highest frequency. This implies that a correct synthesis of input waveform is obtained only when the fn and ζ values of the arithmetic circuit accurately correspond to fn and ζ values of the catheter-manometer system used at the time of output waveform observation.

Therefore, the cause of the decrease in precision of synthesis/estimation with decreasing fn of the catheter-manometer system lies in the fact that the said manometer system has become a complex system (being no longer a secondary system) that cannot be specified by fn and ζ owing to some additive elements, in which instance a decrease in fn invariably occurs [9]. The theory of synthesis/estimation heretofore described will not hold if the catheter-manometer system cannot be dealt with as a secondary system. Such a phenomenon occurs when air bubbles are trapped in the region of a connector such as a three-way tap, when dissolved air gasifies to microbubbles which float locally within the liquid, or for other reasons [9].

The above discussion also means that the method whereby these parameters are calculated from the frequency characteristics measured in a laboratory is liable to involve noticeable errors when applied in the clinical setting; measurements should be made with a catheter kept in-lying. The flushing method and the tapping method have been reported from that Viewpoint [1].

The key to improving the precision of input waveform estimation lies in assembling and maintaining a catheter-manometer system so as to sustain the peak frequency as high as possible. Thereby, the catheter-manometer system can be handled as a true secondary system and, at the same time, the fidelity of output waveform itself is improved. The second and first differential terms in Equation 4 can be considered between-input-and-output error or correction terms. To heighten peak frequency is synonymous with to enlarge ω_n and reduce ζ [3], that is, to reduce the magnitude of the coefficients in the second and first differential terms. In other words, to sustain high peak frequency means to minimize the influence of the secondary system on output waveform.

In conclusion, it is of importance to keep the peak frequency of the catheter-manometer system as high as possible in order to correctly estimate the input waveform from the output waveform obtained with the said manometer system. This is a requisite for obtaining faithful waveforms, but does not mean that estimation of an input waveform is unnecessary if a faithful waveform is obtained. It implies that true intravascular pressure without involvement of the influence of a catheter can be estimated by accurate assemblage of a catheter-manometer system.

REFERENCES

1) Fukuyama T, Kinefuchi Y, Saito S, *et al*: Measuring frequency characteristics of a catheter-manometer system with a pulmonary artery catheter in place. Technology in Anesthesia and Intensive Medical Care – 2001, Kokuseido (Tokyo) pp. 35-39, 2001
2) Graeme JG, Tobey GE, Huelsman LP: Operational amplifiers. Design and applications. McGraw-Hill, 1971
3) Hayashi K, Tanaka Y, Hashimoto S: An attempt at on-line estimation of blood pressure waveform at the actual measurement site, based on invasive arterial pressure monitoring waveforms. Technology in Anesthesia and Intensive Medical Care – 2001, Kokuseido (Tokyo) pp. 21-30, 2001
4) Kinefuchi Y, Suzuki T, Takiguchi M, *et al*: Natural frequency/damping coefficient relationship of catheter-manometer system required for high-fidelity measurement of pulmonary arterial pressure. J Anesthesia 7: 419-426, 1993
5) Maeda K, Kinefuchi Y, Takiguchi M, *et al*: Measurement of effluent calories with a calorimeter in mice. Technology in Anesthesia and Intensive Medical Care – 1994, Kokuseido (Tokyo) pp. 9-12, 1994
6) Milhorn HT: Application of control theory to physiological systems. WB Saunders, London, 1966
7) Savitzky A, Golay MJ: Smoothing and differentiation of data by simplified least squares procedures. Analytical Chemistry 36: 1627-1639, 1964
8) Stephenson FW: RC active filter design handbook. John Wiley & Sons, 1985
9) Suzuki T, Kinefuchi Y, Fukuyama H, *et al*: Practical considerations for improving the dynamic response of the conventional pulmonary catheter manometer system. J J Cir Cont 18: 242-251, 1997