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Validity of the Adaptive Filter for Accurate Measurement of Cardiac Output in Impedance Cardiography

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ONO, T., YASUDA, Y., ITO, T., BARROS, A.K., ISHIDA, K., MIYAMURA, M., YOSHIZAWA, M. and YAMBE, T. *Validity of the Adaptive Filter for Accurate Measurement of Cardiac Output in Impedance Cardiography.* Tohoku J. Exp. Med., 2004, **202** (3), 181-191 ── The purpose of this study was to assess the validity of an adaptive filter, the scaled Fourier linear combiner (SFLC), in the impedance cardiography (ICG). Eight healthy males underwent constant-load bicycle exercise at different intensities from unloaded to near maximal intensity. The stroke volume (SV) and cardiac output ($\dot{Q}$) measured by ICG at each condition were compared with those by the CO_2 rebreathing method. We found that the noises were greatly reduced in the impedance waveform and that the inflection points, so-called the B- and X-points, were clearly detected even during strenuous exercise using the SFLC. Although a high correlation was observed between $\dot{Q}$s measured by the two methods, the mean values of $\dot{Q}$s in each method differed significantly and the regression line also differed significantly from the identity line. Likewise, a significant correlation was observed between SVs obtained by the two methods, but a significant difference in the group mean values and a trend of the regression line were observed. These findings suggest that the use of SFLC in ICG improves the performance in eliminating the noises and in detecting the inflection points in the waveform, thereby contributing to the accurate and beat-to-beat measurements of SV and $\dot{Q}$ especially during exercise.
──── impedance cardiography; adaptive filter; cardiac output; stroke volume; CO_2 rebreathing

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Continuous and noninvasive measurement of hemodynamic variables such as stroke volume (SV), cardiac output ($\dot{Q}$) and other time dependent variables indicating cardiac cycle, are important in assessing cardiovascular adaptation to exercise in sports and exercise physiology. The impedance cardiography (ICG) that was first introduced by Kubicek et al. (1966) has been used to estimate SV and $\dot{Q}$ due to its simple and easy method, however, unreliability and doubt of the measurements still remain. One of the causes of such unreliability may be related to the artifacts derived from respiration, movement, and other unknown sources. Several attempts had hitherto been performed to eliminate the noises mainly derived from respiration in ICG: a narrow band-pass filter around the cardiogenic frequency (Yamamoto et al. 1988), a moving-window technique to identify the breathing artifacts (Eiken and Segerhammar 1988), and a high-pass third-order Butterworth filter with voluntary cardio-respiratory synchronization (Wang et al. 1991; Raza et al. 1992). However, those techniques did not fully mention the noises derived from body movements. Therefore, the most popular tool for data processing in the impedance cardiograph during exercise is an ensemble averaging technique (Muzi et al. 1985; Miyamoto et al. 1988; Yoshida et al. 1993), although this technique loses beat-to-beat measurements. Recently Barros et al. (1995) proposed an adaptive filtering technique named as a scaled Fourier linear combiner (SFLC) that could selectively pass the narrow band signals correlated to the R-R intervals and its harmonics. However, they could show the usefulness of the filter for only mild exercise, and did not practically estimate $\dot{Q}$ and SV during strenuous exercise using such a filtering algorithm.

On the other hand, numerous studies had been conducted to examine the accuracy of ICG on the measurement of $\dot{Q}$ and sometimes of SV by comparing them with other reference methods: a CO_2 rebreathing method, a direct Fick method and a thermodilution method. Most of these studies reported a high and significant relationship between $\dot{Q}$s measured by ICG and other reference methods both at rest and during exercise (Du Quesnay et al. 1987; Miles and Gotshall 1989; Moore et al. 1992). However, despite a high correlation being observed between the methods, recent reviews have cast doubt on the accuracy of the measurement for SV and $\dot{Q}$ using ICG, because of a weak theoretical foundation, an inconsistency in the methodology, and a poor statistical analysis to ascertain the accuracy of the ICG (Smith et al. 1988; Casaburi 1993; Jensen et al. 1995; Warburton et al. 1999).

The aims of the present study are, therefore, to examine 1) the validity for the application of the SFLC on the elimination of noises in the impedance waveform from rest to strenuous exercise, and 2) the accuracy of the present system on the measurement of SV and $\dot{Q}$ by comparing these variables with the CO_2 rebreathing method by using different statistical analyses.

Materials and Methods

Subjects

Eight healthy male subjects volunteered for this study after their informed consent was obtained. The average values and±S.D. of age, height and body mass were 33.3±9.6 years, 173.4 ±2.8 cm, and 67.6±4.8 kg, respectively. Each subject has been familiarized with all measurement devices and the experimental procedure before the study.

Experimental procedure

The subjects underwent bicycle exercises at different intensities. The intensity of exercise was selected from one of two groups, (0, 30, 60, 90, 120, 150 and 180 watts), or (0, 40, 80, 120, 160 and 200 watts) according to the individual's physical fitness. Each experiment was done on different days. Each exercise bout consisted of 1 minute unloaded exercise followed by a chosen intensity of exercise for approximately 7 minutes using a computer-based electro-braked bicycle ergometer designed in our laboratory (Ito et al. 1996). The ICG, ECG, PCG and respiratory variables were measured during the initial 6 minutes including the unloaded exercise for 1 minute, and then the CO_2 rebreathing test followed.

Respiratory variables were measured using a mixing chamber method. The main instruments of the system consist of a pneumotachograph (WFMU-1000, Westron, Kashiwa), a mass spectrometer (WSMR-1400, Westron), and a personal computer (9801F3, NEC, Tokyo). The oxygen uptake ($\dot{V}O_2$), carbon dioxide output ($\dot{V}CO_2$) and minute ventilation ($\dot{V}E$) were continuously measured by 20-second intervals.

Measurement of cardiac output by ICG

Transthoracic impedance was measured by a standard impedance unit (AI-601G, Nihon Kohden, Tokyo), which generates a constant sinusoidal current of 0.35 mA with a frequency of 50 kHz. Four electrically connected spot electrodes were placed around the base of the neck and five identical electrodes placed around the thorax at the level of the xiphoid serves as the inner voltage electrodes. The upper current electrode was set at the central forehead and the lower current ones at the lateral sides of the lower ribcage. The ECG and PCG signals were obtained with a standard method from the chest using a bioamplifier (AC-601G; Nihon Kohden) and a PCG unit (AS-601H, Nihon Kohden), respectively. The impedance signals; basic thoracic impedance (Z_0) and its relative change (ΔZ), and the ECG signal were continuously recorded on a data recorder (RD-120TE, TEAC, Musashino), then re-sampled to a personal computer (PC9821Ap2, NEC) at a frequency of 200 Hz through a 12-bits A/D converter, and finally stored on a MO disk for subsequent analyses.

The dZ/dt signal was obtained from the first derivative of the ΔZ in the ICG, and then the adaptive filter, the scaled Fourier linear combiner (SFLC), was applied on the raw dZ/dt signal. An algorithm of the SFLC was based on the hypothesis that a signal could be expressed as the sum of sine and cosine in a Fourier series in the period T determined by the R-R interval. By using this algorithm, frequency component synchronous with R-R intervals and its harmonics of 16 orders can be selectively picked up from the signal. The algorithm is detailed in our previous study (Barros et al. 1995). The signal processing with SFLC required for about 3 minutes in each serial data.

The SV was calculated from a standard equation proposed by Kubicek et al. (1966), which reads as follows:

$$\mathrm{SV}=\rho(\mathrm{L}/Z_0)^2\,\mathrm{ET}\cdot \mathrm{d}Z/\mathrm{d}t_{\min}$$

where SV is ventricular stroke volume (ml); ρ is blood resistivity and set at 135 cm in this study; L is the shortest distance between the inner series of electrodes at the abdominal side (cm); Z_0 is the basal thoracic impedance (Ω); ET is the ventricular ejection time (s) determined as the duration from points B to X; dZ/d$t_{\min}$ is the most negative deflection of the dZ/dt signal during systole measured from zero line ($\Omega\cdot s^{-1}$). In the present study, we developed an algorithm for detecting the B- and X-points, dZ/d$t_{\min}$ and ET, automatically from the filtered dZ/dt waveform. Cardiac output ($\dot{Q}$) was calculated in beat-to-beat basis by a multiplication of SV and heart rates (HR).

Measurement of cardiac output by CO_2 rebreathing method

Although the CO_2 rebreathing method is not a gold standard for measuring $\dot{Q}$, we used this method as a reference because of its simple and noninvasive way. The subjects breathed 5% CO_2 with O_2 gas from the rebreathing bag. The gas volume in the bag was set approximately 1.5 times larger than the expected tidal volume. The breathing frequency was controlled at 30 breath·min^{-1} during exercise and at 20 breath·min^{-1} at rest by a metronome. During each breath, the subjects inspired almost the whole volume from the bag to ensure rapid mixing. In the application of the CO_2 rebreathing method, the partial pressure of mixed venous CO_2 ($P\bar{v}CO_2$) was estimated from the exponential rise in CO_2 concentration in the lung-bag system during rebreathing by the graphical extrapolation method (Klausen 1965). The arterial CO_2 pressure (P_aCO_2) was estimated from the end-tidal CO_2 (Jernérus et al. 1963). The $P_{\bar{v}}CO_2$ and P_aCO_2 were converted to the content of both arterial (C_aCO_2) and mixed venous blood ($C_{\bar{v}}CO_2$) respectively, through the use of a standard dissociation curve in the oxygenated blood. With that the cardiac output ($\dot{Q}$) was calculated by the Fick equation:

$$\dot{Q} = \dot{V}CO_2 \,/\, (C_{\bar{v}}CO_2 - C_aCO_2)$$

The stroke volume (SV) was calculated from the values of $\dot{Q}$ and HR measured as the average for the last 1 minute of each trial (trial duration from 4 to 5 minutes).

Statistical analysis

Mean values and standard deviations of each variable obtained during the last 1 minute of rest or exercise were calculated as the representative value for each individual at each condition. The $\dot{Q}$ and SV values measured by ICG ($\dot{Q}_{ICG}$ and SV_{ICG}) and by the CO_2 rebreathing method ($\dot{Q}_{REB}$ and SV_{REB}) were compared in the following ways: 1) by a linear regression analysis between $\dot{Q}_{ICG}$ and $\dot{Q}_{REB}$, and between SV_{ICG} and SV_{REB}, 2) by the Bland and Altman's bias analysis (Bland and Altman 1986), and 3) by the paired *t*-test and Wilcoxon signed-ranks test for the difference between $\dot{Q}_{ICG}$ and $\dot{Q}_{REB}$, and between SV_{ICG} and

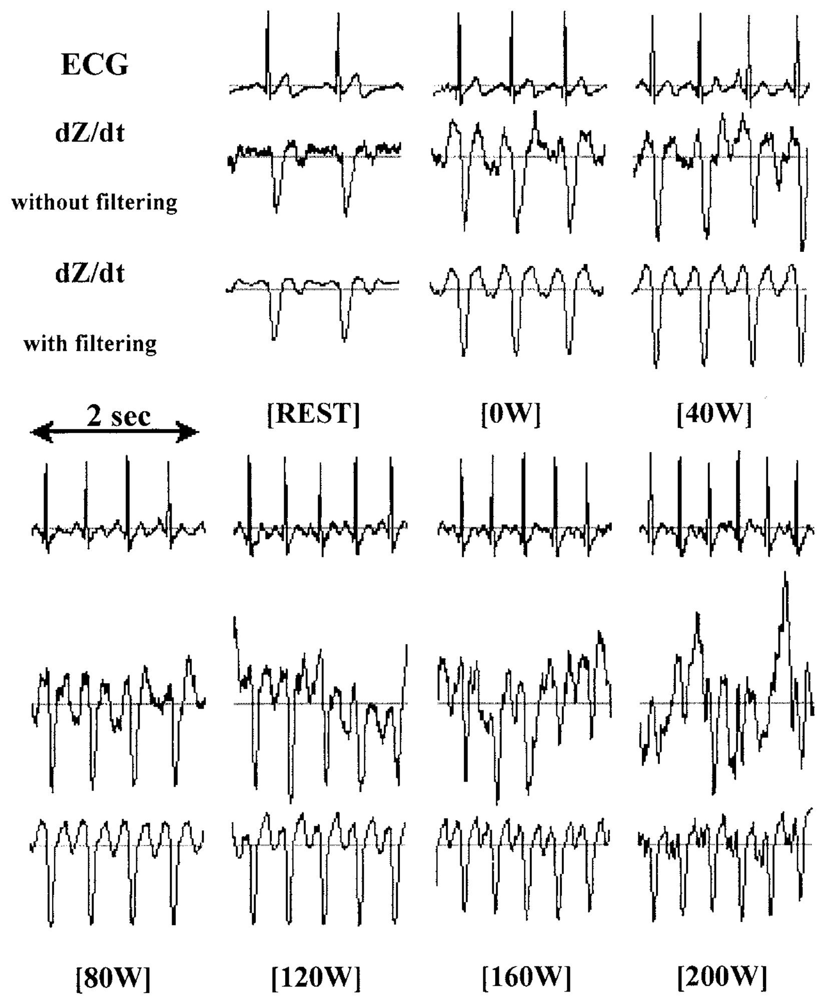

Fig. 1. Schematic recordings of ECG and ICG (impedance derivative recordings: d*Z*/d*t* without filtering, and d*Z*/d*t* with filtering) obtained at rest and during light to near maximum exercises for one subject. Horizontal lines show the zero base of d*Z*/d*t*.

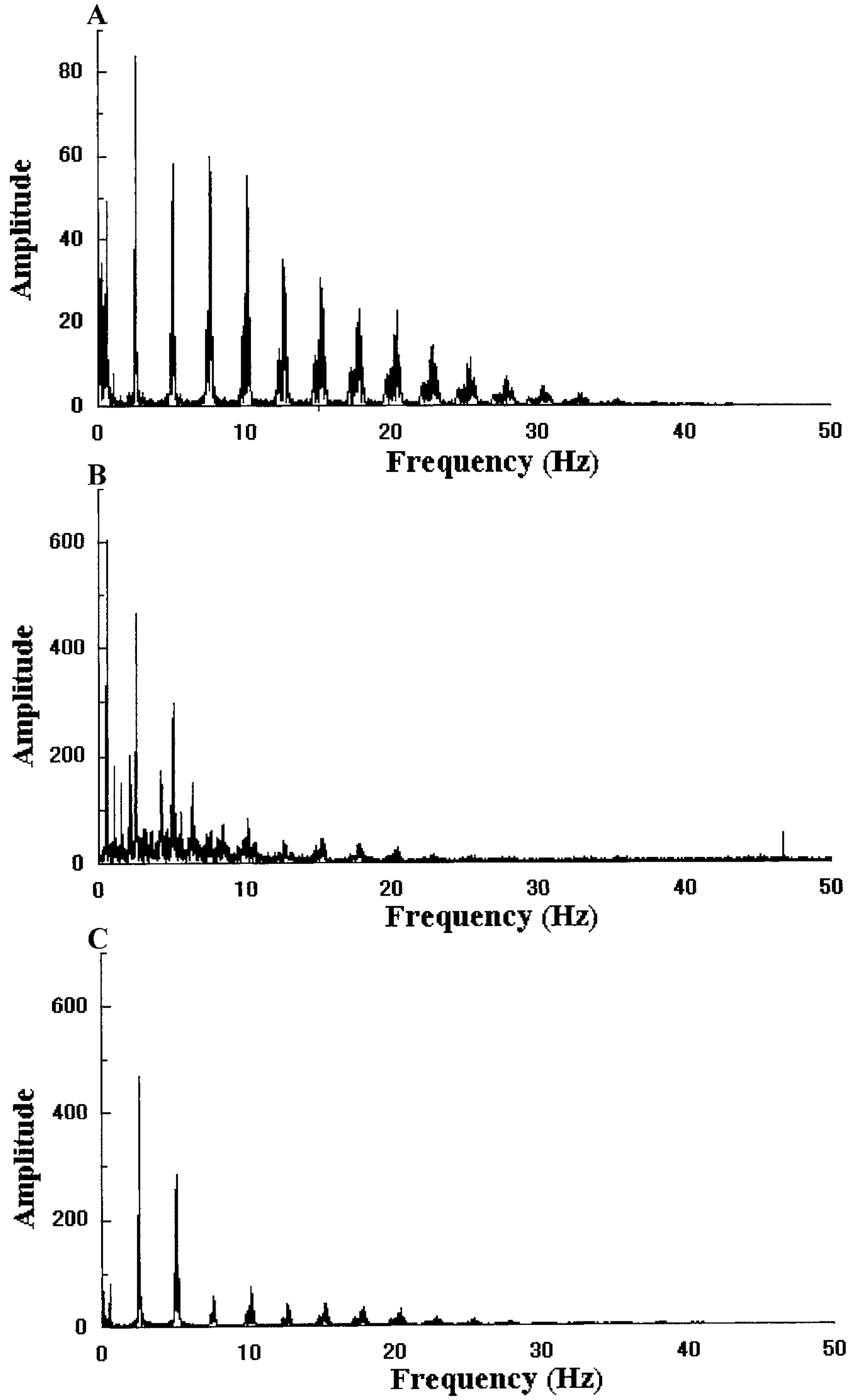

Fig. 2. The power spectrogram of the ECG (A), d*Z*/d*t* without filtering (B) and d*Z*/d*t* with filtering (C) obtained during the last 1-minute of exercise at 160 W.

SV_{REB}. Statistical significance was set at 0.05.

Results

Fig. 1 shows typical examples of the ECG and ICG signals with and without filtering obtained at rest and during exercise at different intensities. In the d*Z*/d*t* waveform, a baseline oscillation and a high frequency noise seemed to be mostly eliminated by filtering even during heavy exercise.

Fig. 2 indicates the power spectra of ECG and d*Z*/d*t* waveforms with and without filtering obtained during the last 1-minute of exercise at 160 W for the same subject in Fig. 1. It was clearly demonstrated that the harmonic components synchronous with R-R intervals in the d*Z*/d*t* waveform remained, and the asynchronous components with R-R intervals were mostly eliminated.

Fig. 3 shows a typical example for the detection of the B- and X-points and concomitant changes of ECG and PCG obtained at moderate exercise. The B-point seemed to appear in accordance with the onset or just after the first heart sound (S_I), and the X-point just at the onset of the second heart sound (S_{II}) in the PCG, respectively. However, it was frequently difficult to identify the B- and the X-points from the raw d*Z*/d*t*

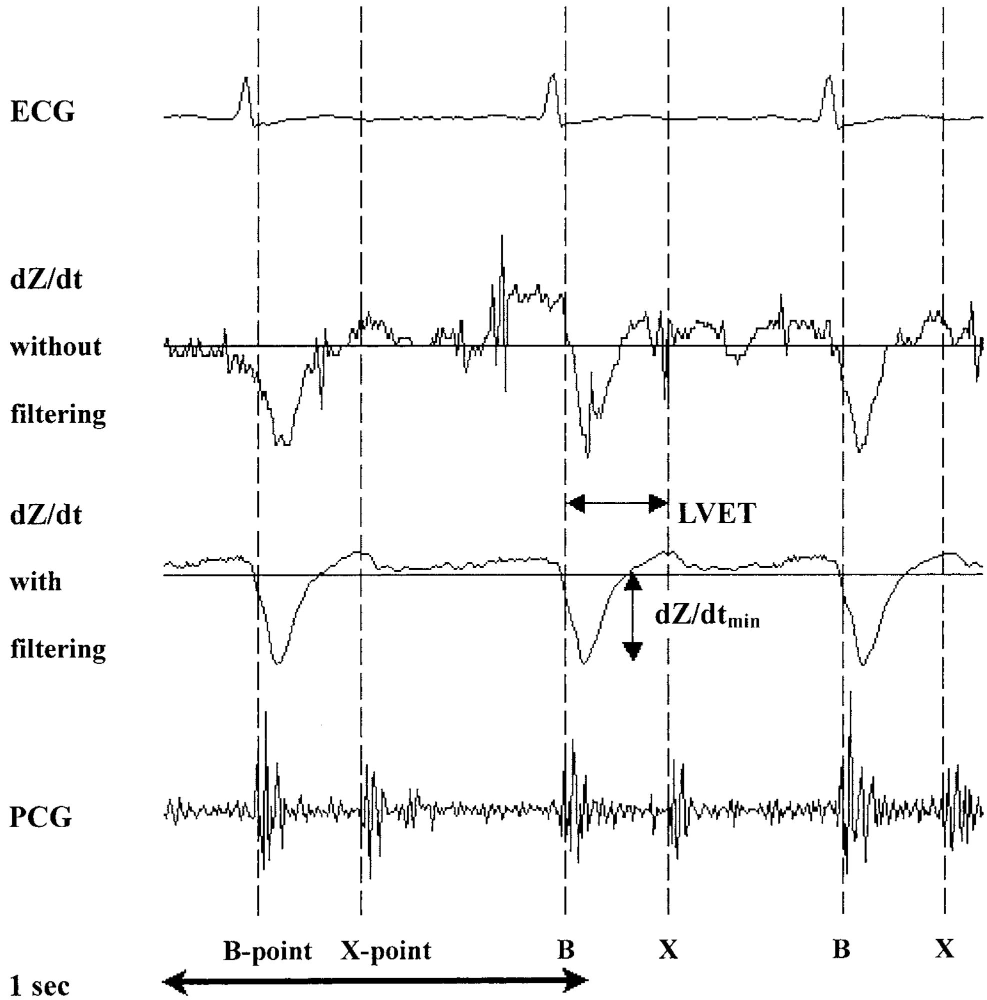

Fig. 3. Schematic recording of ECG, ICG (impedance derivative recordings: d*Z*/d*t* without filtering, and d*Z*/d*t* with filtering) and PCG obtained during moderate exercise (120 W) for one subject. Horizontal lines show the zero base of d*Z*/d*t* and the vertical dotted lines show the B- and X-points of d*Z*/d*t*, respectively.

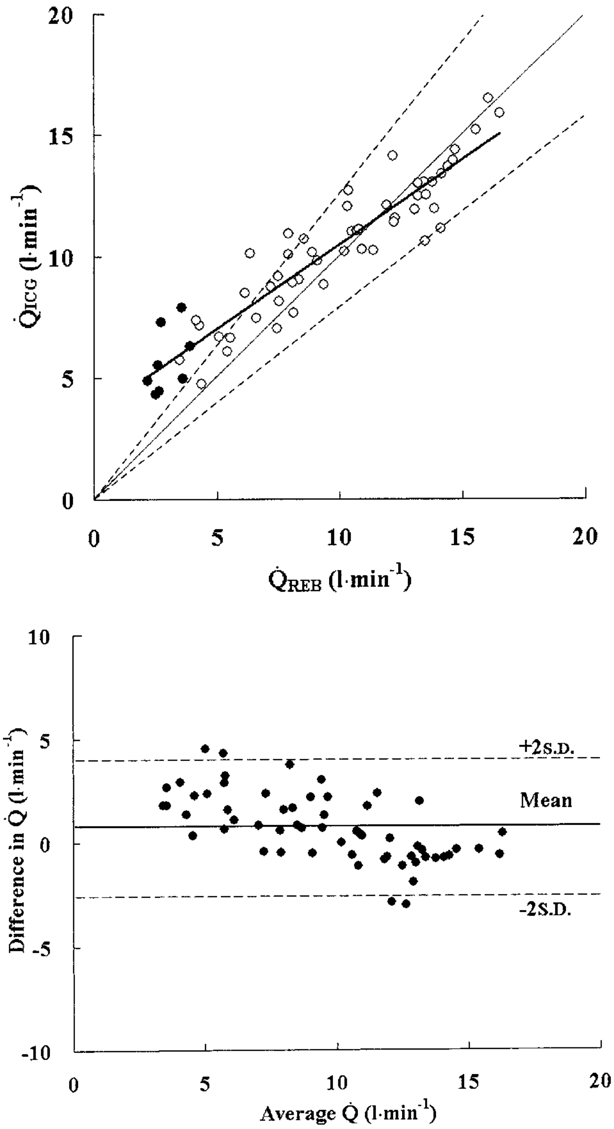

Fig. 4. Upper panel: Scatterplot of $\dot{Q}$ calculated by ICG ($\dot{Q}_{ICG}$) vs. CO_2 rebreathing method ($\dot{Q}_{REB}$) obtained from 8 subjects at rest (black circles) and during exercise (white circles). The regression equation and statistical data are Y=0.701X+3.49 (r=0.936, p<0.001). The line of identity and the lines±20% are also indicated.
Lower panel: Bland and Altman plot showing the relation between $\dot{Q}_{ICG}$ and $\dot{Q}_{REB}$. Difference between measurements of cardiac output ($\dot{Q}_{ICG}$ $-\dot{Q}_{REB}$), as determined using impedance cardiography and CO_2 rebreathing method *vs.* the average of the two values; the mean difference (horizontal line) and 95% limits of agreement (dotted lines) are also shown. The mean difference±S.D. in $\dot{Q}$ was 0.727±1.636 l·min^{-1}.

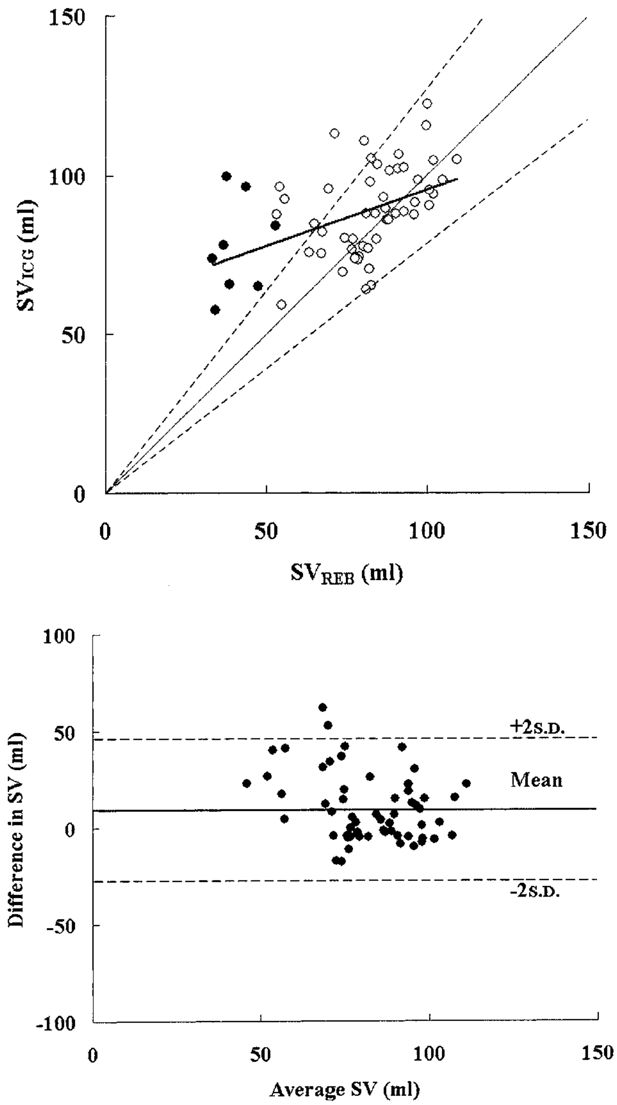

Fig. 5. Upper panel: Scatterplot of SV calculated by ICG (SV_{ICG}) vs. CO_2 rebreathing method (SV_{REB}) obtained from 8 subjects at rest (black circles) and during exercise (white circles). The regression equation and statistical data are Y=0.361X+59.6 (r=0.477, p<0.001). The line of identity and the lines±20% are also indicated.
Lower panel: Bland and Altman plot showing the relation between SV_{ICG} and SV_{REB}. Difference between measurements of stroke volume (SV_{ICG} – SV_{REB}), as determined using impedance cardiography and CO_2 rebreathing method *vs.* the average of the two values; the mean difference (horizontal line) and 95% limits of agreement (dotted lines) are also shown. The mean difference±S.D. in SV was 10.3±17.9 ml·min^{-1}.

waveform especially during exercise as shown in the figure. These figures (Figs. 1, 2 and 3) could show the validity of using SFLC in eliminating noises and in detecting so-called B- and X-points precisely during exercise.

A scatterplot of $\dot{Q}$s measured by ICG ($\dot{Q}_{ICG}$) and the CO_2 rebreathing method ($\dot{Q}_{REB}$), and a biasplot of $\dot{Q}$s are illustrated in Fig. 4. It should be mentioned that a small time delay existed between the measurements of $\dot{Q}_{ICG}$ and $\dot{Q}_{REB}$. A high and significant (r=0.936, p<0.001) correlation was observed between them, but statistical differences between $\dot{Q}_{ICG}$ and $\dot{Q}_{REB}$ (t=3.37, df=58, p<0.01 by paired t-test, and z=2.91, N=59, p<0.01 by Wilcoxon signed-ranks test) were also observed. Furthermore the slope coefficient and the intercept of the regression line between $\dot{Q}_{ICG}$ and $\dot{Q}_{REB}$ were significantly different from unity and zero, respectively. A biasplot also showed that 4 out of 59 points of data of $\dot{Q}$s were outside of the 95% limit of confidence.

A scatterplot of the SVs measured by ICG (SV_{ICG}) and the CO_2-rebreathing method (SV_{REB}) and the precision of the bias plotting for SV are illustrated in Fig. 5. Although the regression analysis showed a significant relationship (r=0.477, p<0.001) between SV_{ICG} and SV_{REB}, significant differences (t=4.38, df=58, p<0.001 by paired t-test, and z=3.51, N=59, p<0.001 by Wilcoxon signed-ranks test) were observed. Furthermore, significant differences in the slope coefficient and the intercept of the regression equation from unity and zero were also observed. A bias plot of all SV revealed that 2 out of 59 data points were outside of the confidence level.

Discussion

Filtering performance

In the present study, it seems in the dZ/dt waveform, that a high-frequency vibration and a low-frequency base line oscillation was mostly eliminated (Fig. 1), and that asynchronous components with ECG could be selectively eliminated (Fig. 2) by using the SFLC. Thereby the detection of the inflection points corresponding to the B- and X-points in the dZ/dt waveform became clearer even during strenuous exercise (Fig. 3).

It has been reported, regarding the ICG measurements, that breathing or movement artifacts cause distortion and may restrict a wide and accurate measurement of the system especially during exercise. Several technical attempts: a narrow band-pass filter around the cardiogenic frequency (Yamamoto et al. 1988); a moving-window technique to identify the breathing artifacts (Eiken and Segerhammar 1988); and a high-pass third-order Butterworth filter with voluntary cardiorespiratory synchronization (Wang et al. 1991; Raza et al. 1992), had been developed to minimize the noises. However, these techniques were mainly aimed to eliminate the noises caused by respiration and might not be sufficiently considered for movement-derived artifacts. Movement-derived artifacts were enlarged with increasing exercise intensity as shown in Fig. 1, and located in a wide frequency band. Therefore the elimination of the movement-derived artifact is important in the ICG measurement especially during strenuous exercise. From this point of view, the SFLC is thought to be superior to other techniques for noise elimination in ICG. However, it should be mentioned that some problems still remain in the application of this technique. First, the SFLC has an essential problem in which noises cannot be eliminated whenever the frequency between movement and cardiac rhythms are synchronized or integrally harmonized. For example when the pedaling frequency is 60 rpm and HR is around 120 beats·min^{-1}, the movement-derived artifact may overlap at the frequency of HR. In this case, noises caused by body movement might not be eliminated. Second, when using the SFLC, the order of harmonics should be set to cover all of the frequency bands of the signal. In the present study, the order of harmonics was set at 16. When the frequency of R-R intervals corresponded to around 2.5 Hz, as indicated in Fig. 2, high frequency components exceeding 40 Hz in the dZ/dt waveform would be eliminated. Cutting-off the high frequency components will affect on the determination of the dZ/dt_{min}, but in the present study, the effect of order of harmonics was not examined on the determination of the dZ/dt_{min}. Although these problems still remain, the SFLC would be a useful tool for eliminating noises and the beat-to-beat measurement of SV and $\dot{Q}$ by ICG, especially during exercise.

Accuracy of the measurement of SV and $\dot{Q}$

In the present study, a high and significant correlation (r=0.936, p<0.001) was observed between $\dot{Q}$s measured by ICG and the CO_2 rebreathing method in overall measurements from rest to heavy exercise, while significant differences were identified by the paired t-test and the Wilcoxon signed-ranks test between them. The bias analysis also revealed that 4 of 59 overall plots were out of the 95 % limit of confidence. Furthermore the slope coefficient and the intercept of the regression equation between $\dot{Q}_{ICG}$ and $\dot{Q}_{REB}$ differed significantly from unity and zero, respectively. With respect to the SV, a low but significant relationship (r=0.477, p<0.001) was found between SV_{ICG} and SV_{REB}, but a significant difference and a significant trend in the regression equation between them were also observed as being the same in $\dot{Q}$.

Up to now, numerous studies had been performed to examine the accuracy of ICG on the measurement of $\dot{Q}$ and sometimes SV by comparing them with other reference standards including a direct Fick, a thermodilution, a dye-dilution and a CO_2 rebreathing methods. The overall correlation coefficients (r) of $\dot{Q}$s between ICG and the reference methods were reported as ranging from 0.50 to 0.98 (Miles and Gotshall 1989; Jensen et al. 1995; Warburton et al. 1999). With respect to the relationship of $\dot{Q}$s between ICG and the CO_2 rebreathing method, similar values of r from 0.56 to 0.95 were also observed. From such studies, it has been believed that the ICG is a reliable and valid tool for measuring $\dot{Q}$ not only at rest but also during exercise (Teo et al. 1985; Du Quesnay et al. 1987; Moore et al. 1992). The grounds for such reliability of the ICG relied upon the high correlation coefficient of $\dot{Q}$s between ICG and the reference methods. Smith et al. (1988) suggested that a high correlation coefficient in $\dot{Q}$s measured by different methods from rest to exercise might depend on a linear increase of HR, and might not indicate the validity of ICG. Furthermore, they suggested that the lower correlation coefficient in SV compared to that in $\dot{Q}$ between the different methods might support above hypothesis. Jensen et al. (1995) claimed the necessity for applying a bias analysis or a mean difference analysis to compare the data obtained by different methods. Recent studies with regard to the accuracy of ICG have used other statistical analyses with the correlation analysis (Teo et al. 1985; Moore et al. 1992; Bogaard et al. 1997). For example, Teo et al. (1985) and Bogaard et al. (1997) applied a linear regression analysis to ascertain whether a significant variation existed in the regression equation between $\dot{Q}$s. However, a disagreement was seen in their results. Teo et al. (1985) who used the direct Fick method as a reference, reported no statistical difference in the slope coefficient and intercept of the regression equation from the identity line, but Bogaard et al. (1997) reported a systematic difference in both the slope coefficient and the intercept. The present results agreed with the latter study, because the slope coefficient and the intercept of the regression equation differ significantly from unity and zero, respectively, as shown in Fig. 4. Bogaard et al. (1997) also applied the mean difference analysis of the groups and the bias analysis on the two measurements, and noted no systematic difference and bias on the overall data obtained both at rest and during exercise in $\dot{Q}$s. In contrast, the present results show a systematic difference in $\dot{Q}$s between the two methods, and four out of 59 data points were outside of the 95 % limit of confidence.

With respect to SV, Bogaard et al. (1997) reported a significant correlation (r=0.79) between SVs measured by ICG and the CO_2 rebreathing method, but significant differences in the slope coefficient and intercept of the regression equation between them. Smith et al. (1988) reported a poor but significant relationship (r=0.24) in the stroke volume index instead of SV between the ICG and the CO_2 rebreathing method. The present results in SV show an intermediate value of the two previous studies noted above. At any rate, it is unquestionable that the agreement in SV between ICG and the reference methods were reduced compared to that of $\dot{Q}$.

According to the previous studies and the present one, it may be concluded that the accuracy of ICG for the measurement of SV and $\dot{Q}$ are still questionable. The real reasons of the discrepancies among the results cannot be resolved here. Any methodological problems; the calculation formula, the type and position of the electrodes, the measurement of the length between the inner electrodes, and the estimation of blood resistivity and etc., may be involved. Although the ICG still has uncertainties in its accuracy for measuring SV or $\dot{Q}$, there is no other similar equipment or methods that can measure SV and $\dot{Q}$ with continuous, noninvasive, and in beat-to-beat basis even during heavy exercise. Furthermore, the ICG can simultaneously provide other variables such as the left ventricular ejection time and the pre-ejection period. These variables have also been used as good indices for assessing the contractility of the heart (Máttar et al. 1991). Further studies would be, therefore, required for the accurate and wide application of the ICG.

Conclusion

In the present study, the validity of the adaptive filter (SFLC) that could selectively pass the harmonic components of the R-R intervals was assessed in the elimination of noises and in the accurate measurements of SV and $\dot{Q}$ from rest to strenuous exercise. With using the SFLC, the inflection points of so-called B- and X-points in the dZ/dt waveform could be detected more precisely, and consequently, estimation errors in the dZ/dt_{min} and the LVET could be greatly reduced. Comparisons of $\dot{Q}$s measured by ICG and the CO_2 rebreathing method both at rest and during exercise at different intensities revealed a high and significant correlation (r=0.936) between them. However, a statistical difference in the group mean value and a significant trend of the regression equation were also detected. Furthermore, the relationship between SVs measured by the two methods showed a lower agreement rather than that in $\dot{Q}$. From these results, it is suggested that the SFLC may improve the performance of ICG especially in the beat-to-beat measurement of the dZ/dt_{min} and the ET, however, unreliability still remained in the accuracy of $\dot{Q}$ and SV. Other factors, rather than the noises in the impedance signal, might be involved in the accuracy of measurements. Further studies will be required.

References

Barros, A.K., Yoshizawa, M. & Yasuda, Y. (1995) Filtering noncorrelated noise in impedance cardiography. *IEEE Trans. Biomed. Eng.*, **42**, 324-327.

Bland, J.M. & Altman, D.G. (1986) Statistical methods for assessing agreement between two methods of clinical measurement. *Lancet*, **1**, 307-310.

Bogaard, H.J., Woltjer, H.H., Postmus, P.E. & de Vries, P.M.J.M. (1997) Assessment of haemodynamic response to exercise by means of electrical impedance cardiography: method, validation and clinical applications. *Physiol. Meas.*, **18**, 95-105.

Casaburi, R. (1993) Evaluation of cardiac output by thoracic electrical bioimpedance during exercise in normal subjects. *Chest*, **104**, 985.

Du Quesnay, M.C., Satoute, G.J. & Hughson, R.L. (1987) Cardiac output in exercise by impedance cardiography during breath holding and normal breathing. *J. Appl. Physiol.*, **62**, 101-107.

Eiken, O. & Segerhammar, P. (1996) Elimination of breathing artifacts from impedance cardiograms at rest and during exercise. *Med. Biol. Eng. Comput.*, **26**, 13-16.

Ito, T., Ono, T., Yasuda, Y., Nishioka, M. & Horiuchi, O. (1996) Assessment for the equivalent inertia and energy transfer loss of the electro-magnetic bicycle ergometer. *Jpn. J. Sports Sci.*, **15**, 127-133. (in Japanese with English abstract)

Jensen, L., Yakimets, J. & Teo, K.K. (1995) A review of impedance cardiography. *Heart Lung*, **24**, 183-193.

Jernérus, R., Lundin, G. & Thomson, D. (1963) Cardiac output in healthy subjects determined with a CO_2 rebreathing method. *Acta Physiol. Scand.*, **59**, 390-399.

Klausen, K. (1965) Comparison of CO_2 rebreathing and acetylene methods for cardiac output. *J. Appl. Physiol.*, **20**, 763-766.

Kubicek, W.G., Karnegis, J.N., Patterson, R.P., Witsoe, D.A. & Mattson, R.H. (1966) Development and evaluation of an impedance cardiac output system. *Aerospace Med.*, **37**, 1208-1212.

Máttar, J.A., Shoemaker, W.C., Diament, D., Lomar, A., Lopes, A.C., Freitas, E.D., Stella, F.P. & Factore, L.A.P. (1991) Systolic and diastolic time intervals in the critically ill patients. *Crit. Care Med.*, **19**, 1382-1386.

Miles, D.S. & Gotshall, R.W. (1989) Impedance cardiography: noninvasive assessment of human central hemodynamics at rest and during exercise. *Exerc. Sport Sci. Rev.*, **17**, 231-263.

Miyamoto, Y., Kawahara, K., Nakazono, Y., Grucza, R., Sugawara, T. & Sato, K. (1988) The origin of the initial abrupt increase in ventilation at the onset of muscular exercise (phase 1) in man. *Tohoku J. Exp. Med.*, **156**, Suppl., 113-123.

Moore, R., Sansores, R., Guimond, V. & Abboud, R. (1992) Evaluation of cardiac output by thoracic electrical bioimpedance during exercise in normal subjects. *Chest*, **102**, 448-455.

Muzi, M., Ebert, T.J., Tristani, F.E., Jeutter, D.C., Barney, J.A. & Smith, J.J. (1985) Determination of cardiac output using ensemble-averaged impedance cardiograms. *J. Appl. Physiol.*, **58**, 200-205.

Raza, S.B., Patterson, R.P. & Wang, L. (1992) Filtering respiration and low-frequency movement artifacts from the cardiogenic signal. *Med. Biol. Eng. Comput.*, **30**, 556-561.

Smith, S.A., Russell, A.E., West, M.J. & Chalmers, J. (1988) Automated non-invasive measurement of cardiac output: comparison of electrical bioimpedance and carbon dioxide rebreathing techniques. *Br. Heart J.*, **59**, 292-298.

Teo, K.K., Hetherington, M.D., Haennel, R.G., Greenwood, P.V., Rossall, R.E. & Kappagoda, T. (1985) Cardiac output measured by impedance cardiography during maximal exercise tests. *Cardiovasc. Res.*, **19**, 737-743.

Wang, L., Patterson, R.P. & Raza, S.B. (1991) Respiratory effects on cardiac related impedance indices measured under voluntary cardio-respiratory synchronisation (VCRS). *Med. Biol. Eng. Comput.*, **29**, 505-510.

Warburton, D.E.R., Haykowsky, M.J.F., Quinney, H.A., Humen, D.P. & Teo, K.K. (1999) Reliability and validity of measures of cardiac output during incremental to maximal aerobic exercise. Part II: Novel techniques and new advances. *Sports Med.*, **27**, 241-260.

Yamamoto, Y., Mokushi, K., Tamura, S., Mutoh, Y., Miyashita, M. & Hamamoto, H. (1988) Design and implementation of a digital filter for beat-by-beat impedance cardiography. *IEEE Trans. Biomed. Eng.*, **35**, 1086-1090.

Yoshida, T., Yamamoto, K. & Udo, M. (1993) Relationship between cardiac output and oxygen uptake at the onset of exercise. *Eur. J. Appl. Physiol.*, **66**, 155-160.