

INFLUENCE OF SIGNAL FIDELITY ON IMPEDANCE CARDIOGRAPHICALLY-DERIVED VALUES AT RESTING AND ACCELERATED HEART RATES

Barry E. Hurwitz, Liang-Yu Shyu, Chih-Cheng Lu, Sridhar P. Reddy,
Neil Schneiderman, and Joachim H. Nagel

Behavioral Medicine Research Center, Departments of Psychology &
Biomedical Engineering, University of Miami, Coral Gables, FL 33124

ABSTRACT

Since the spectrum of the impedance cardiogram (ICG) extends from DC to 50 Hz, any amplifier with an upper band limit less than 50 Hz can be expected to produce attenuation and distortion of the ICG. This signal attenuation may be systematically enhanced under conditions of high heart rates (HR) when a greater proportion of signal energy will be in the upper frequency range of the ICG spectrum. Therefore, the present study was designed to assess the influence of amplifier bandwidth and signal fidelity on dZ/dt_{max} , stroke volume (SV), and systolic time intervals (LVET, PEP, QZ, HI). The performance of commonly available commercial systems was tested over a broad range of HRs. The results demonstrated that a digitally differentiated dZ/dt signal using a differentiator with a corner frequency of 50 Hz, when compared with the 15 Hz corner frequency used in the commercial impedance cardiograph, systematically enhanced the dZ/dt_{max} amplitude and SV measurements as HR increased. For SV the increase ranged from 17 to 30% as HR increased from 70 to 150 bpm. Moreover, the digitally filtered signal had greater resolution and produced less prolonged PEP and QZ intervals and greater HI with increasing HR. These findings indicate that impedance cardiographs with insufficient upper bandlimits will differentially influence ICG-derived measurements as HR varies.

INTRODUCTION

Impedance cardiography is a noninvasive, inexpensive, atraumatic, and relatively unobtrusive technique for deriving stroke volume (SV), cardiac output, systolic time intervals and several related cardiovascular parameters. Thus this technique affords the opportunity to perform a comprehensive cardiovascular functional analysis in the resting and behaving human. The value of impedance cardiography for studying cardiovascular regulation in response to dynamic challenge [1] or for clinical diagnostic assessment [2] depends ultimately upon the signal fidelity of the front-end instrumentation.

SPECTRUM OF THE IMPEDANCE CARDIOGRAM

To establish the appropriate signal fidelity requirements for the instrumentation used, it is first necessary to determine the ICG spectrum, which has not previously been documented. An impedance cardiograph with a 2 mA, 100 kHz constant current source and a wideband 265 Hz anti-aliasing filter at the output was used to collect the ICG signal (ΔZ). The ICG spectra of ten continuous cardiac cycles at low (70-89 bpm) and high (130-149 bpm) heart rates (HR) were obtained with a spectral resolution of 0.98 Hz. The spectra were then averaged across the 7 subjects at both HR levels. The purpose of this spectral analysis was to specify the bandwidth, noise and dynamic range necessary to provide optimum performance of an ICG amplifier for subsequent event detection and amplitude measurement. Therefore, the upper bandlimit was defined as the frequency where the signal power drops below the power of the noise (i.e., $SNR < 1$). In Figure 1 (left panel) inspection of the ICG spectrum suggests that the noise component can clearly be identified as the spectrum baseline. This procedure, however, does not take into account the inherent variation of physiological signals. Since detection algorithms generally are based on the typical appearance of these signals, interindividual variance of signal shape may be considered as a measure of the "physiological noise".

Thus the ICG bandwidth was determined by calculating the average spectra, subtracting the baseline noise and deriving the SNR. The SNR for a given frequency was calculated as the ratio of average power and its standard deviation. Then the upper bandlimit may be determined by locating the frequency where SNR declines below one. This procedure limits the spectral band of the signal to those frequencies where the contribution of the true signal is larger than that of noise and interferences. Inspection of Figure 1 (right panel) reveals that the ICG spectrum ranges within 30 Hz at slower HR levels and increases to 50 Hz at faster HR levels, when SNR equals one.

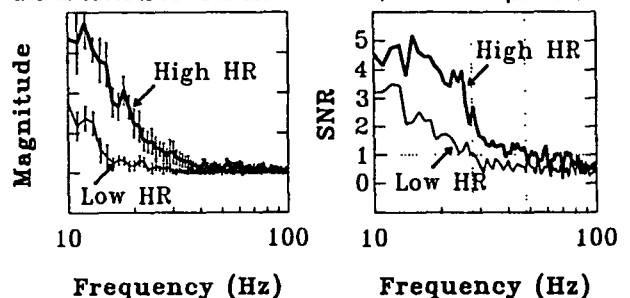


FIGURE 1. Depicted at low (70-89 bpm) and high (130-149 bpm) HR are the mean \pm SD (left) and SNR (right) of the ICG spectrum.

In general, the bandwidth of the commercially available impedance cardiographs appears to be insufficient to produce the ICG- dZ/dt without attenuation or distortion. Since the Minnesota Impedance Cardiograph (MIC) is the most widely used impedance cardiograph, we have selected it to use in this study. A frequency response analysis of the MIC showed upper band limits of 60 Hz for the ΔZ channel and 15 Hz for the dZ/dt channel. Since the ICG spectrum extends to 50 Hz, at the very least ICG systems with corner frequencies less than 50 Hz would be expected to attenuate the signals under conditions of high HRs when a greater proportion of signal energy is in the higher frequency range of the ICG spectrum. Since a HR dependent outcome was anticipated, this study examined the influence of signal fidelity on ICG-derived dZ/dt_{max} , SV and systolic time interval indices over a broad range of HR.

METHOD

Seven healthy men reporting no cardiopulmonary or other medical disorders, aged between 21 and 45 yrs, served as subjects. During the testing session electrocardiogram (ECG), phonocardiogram (PCG) and the ICG signals - ΔZ , dZ/dt and mean thoracic impedance (Z_0) - were recorded using a Grass polygraph and the MIC (model 304B). Physiological measurements were obtained in 30-s samples during rest and during a bicycle exercise procedure in which the workload was adjusted to produce HRs in the following bpm ranges: 70-89, 90-109, 110-129, and 130-149. A standard lead II ECG configuration was used. The PCG was recorded by placing the phonotransducer (Hewlett Packard 21050A) on the cardiac window in the second intercostal space just left of the sternum. This measure provided confirmation of the location of the ICG-X wave reflecting aortic valve closure. The ICG was derived using a tetrapolar electrode configuration, with bands placed 360° around the body [cf. 3]. The mean of the front and back distance between

leads 2 and 3 was measured for later use in calculation of SV using the Kubicek equation [4]. All signals were sampled by an IBM PS/2 Model 70 computer at 1 kHz using an A/D converter (DT 2901). Impedance calibration signals for the three ICG measures were also stored in the computer for later conversion of the ICG measurements to the corresponding units.

The ECG, PCG and dZ/dt signals were displayed simultaneously on a computer display on a beat-by-beat basis. A computed dZ/dt was also displayed, wherein the derivative of the measured ΔZ was calculated using a nonrecursive digital differentiator, with a corner frequency of 50 Hz, transition bandwidth of 5 Hz and minimum stopband rejection of -20 dB. ICG events were located automatically and the displayed event markers could be manually adjusted. The dZ/dtmax was defined as the amplitude difference between B-point and the maximum of the dZ/dt. Left ventricular ejection time (LVET) was determined as the interval between the B point and the X minimum. The mean Z_0 during systole for each cardiac cycle and the blood resistivity constant of 135 $\Omega \cdot \text{cm}$ were used in the Kubicek equation. Systolic time intervals such as pre-ejection period (PEP), and time to maximal ejection velocity (QZ) as well as the contractility index, Heather index (HI), were also measured. The parameters were scored on a beat-by-beat basis and values were averaged over ten consecutive cardiac cycles within each of the four HR ranges.

RESULTS

The data analysis used a repeated measures analysis of variance to compare across the four HR intervals the cardiovascular parameters (dZ/dtmax, SV, LVET, PEP, QZ and HI) derived from the two measurement methods: the dZ/dt measured from the MIC, with a bandwidth of 15 Hz, and the dZ/dt computed from the measured ΔZ by a digital filter with a bandwidth of 50 Hz. Table 1 depicts the mean \pm SD of the dZ/dtmax and SV for each method during the HR intervals. For both dZ/dtmax and SV significant differences between measured and computed methods were observed, with greatest mean differences appearing at the fastest HRs and reducing at the slowest HRs ($p < .005$). However, at each HR level the dZ/dt max and the SV values derived from computed dZ/dt were greater than values derived from measured dZ/dt (see Table 1 for p values). For example, the underestimation of SV by the measured dZ/dt method ranged from 17% to greater than 30% over the HR intervals.

Table 1. Mean (\pm SD) Ejection Velocity (dZ/dtmax) and Stroke Volume (SV) Comparing Measured dZ/dt and Computed dZ/dt
Heart Rate (bpm)

Parameter	Method	70-89	90-109	110-129	130-149
dZ/dtmax (Ω/s)	measured	2.0 \pm 0.4	2.9 \pm 0.4	3.0 \pm 0.4	3.1 \pm 0.5
	computed	2.3 \pm 0.6 [‡]	3.4 \pm 0.4*	3.8 \pm 0.7*	4.1 \pm 0.7*
SV (ml)	measured	67.3 \pm 27	79.2 \pm 21	74.9 \pm 21	67.5 \pm 16
	computed	78.6 \pm 34 [™]	94.4 \pm 28 [‡]	94.8 \pm 30 [‡]	87.9 \pm 23 [‡]

™p < .05; ‡p < .005; *p < .0005 (p values for measured vs. computed)

Figure 2 displays, in a representative subject, greater dZ/dtmax obtained using the computed dZ/dt method (thick line) than the dZ/dtmax measured from MIC (thin line); this difference is present at

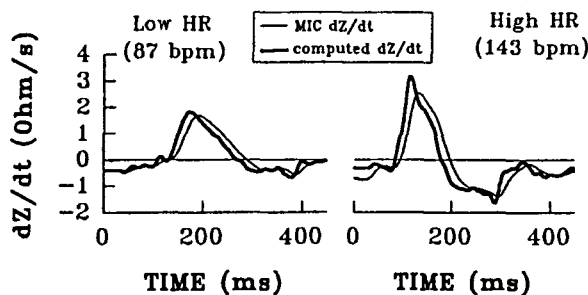


FIGURE 2. Single cycle records at low and high heart rate from a single subject comparing measured MIC dZ/dt with computed dZ/dt.

both low (87 bpm) and high (143 bpm) HRs. Note the greater resolution of the relevant signal events (B point, dZ/dt peak and X wave) in the computed dZ/dt tracings.

Table 2. Mean (\pm SD) Systolic Time Intervals Comparing Measured dZ/dt and Computed dZ/dt

Parameter	Method	Heart Rate (bpm)			
		70-89	90-109	110-129	130-149
LVET (ms)	measured	270.5 \pm 16	227.9 \pm 16	208.6 \pm 12	184.4 \pm 15
	computed	272.7 \pm 15	227.0 \pm 17	207.4 \pm 13	182.7 \pm 16
PEP (ms)	measured	92.7 \pm 24	59.7 \pm 4	57.8 \pm 3	56.9 \pm 6
	computed	80.9 \pm 24*	51.3 \pm 4*	48.8 \pm 4*	47.8 \pm 6*
QZ (ms)	measured	155.5 \pm 22	115.3 \pm 9	108.3 \pm 6	101.7 \pm 6
	computed	131.6 \pm 22*	91.0 \pm 6*	84.5 \pm 5*	78.5 \pm 3*
HI (Ω/s^2)	measured	13.6 \pm 5	24.8 \pm 3	27.9 \pm 5	30.8 \pm 6
	computed	18.9 \pm 9 [™]	37.5 \pm 5*	45.6 \pm 10*	52.5 \pm 10*

™p < .05; ‡p < .005; *p < .0005 (p values for measured vs. computed)

Table 2 depicts the effect of the two measurement methods on the systolic time intervals LVET, PEP, QZ and HI. For HI and QZ a significant increasing difference between methods was found with increasing HR level ($p < .0005$). However, for PEP decreasing difference between methods was obtained with faster HRs ($p < .0005$). Although there was no effect on LVET interval, for PEP, QZ and HI, the computed method resulted in significantly different values than the measured method (see Table 2 for p values).

DISCUSSION

The primary objective of the present study was to assess the influence of signal fidelity on cardiovascular indices by comparing measurements derived from dZ/dt with measurements obtained from the computed derivative of ΔZ at HRs varying over a broad range. The results showed that the 15 Hz low pass filter used in the MIC substantially attenuated the dZ/dt amplitude and SV measurements. This effect occurred both at slower HRs and also to a greater extent at faster HRs, wherein greater than 30% difference in dZ/dtmax was found. The differential influence of signal fidelity may, in part, explain the low correlations, reported in a recent validity study, between ICG and nuclear ventriculography SV estimates during exercise when HRs were increased but not during rest [5].

While there was no influence of signal fidelity on LVET, the enhancement of waveform topography when dZ/dt was computed resulted in less latency of the ICG signal events relative to the ECG-Q wave. Consequently, PEP and QZ intervals were shorter; the larger dZ/dtmax and shorter QZ intervals resulted in greater HI. In summary, these findings indicate that, when an upper bandlimit less than 50 Hz is used to filter dZ/dt an inherent underestimation of dZ/dtmax, SV and HI, and an overestimation of the systolic time intervals, PEP and QZ, occurs. Moreover, the error systematically increases as HR accelerates from resting levels.

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