Reprinted from AMERICAN HEART JOURNAL, St. Louis
Vol 116, No. 2, Pt. 2, Pp. 657-664. August, 1988, (Pruned in the U.S.A.I

A practical, cost-effective, noninvasive system for cardiac output and hemodynamic analysis

Impedance cardiography is a relatively Inexpensive, noninvasive technique for measuring cardiac output on the basis of resistive changes in the thorax to electrical current flow. In conjunction with blood pressure monitoring and physiologic maneuvers, the technique may b. used to monitor thoracic and total body fluid volume and express a variety of contractility indexes, as will as relative and absolute measurements of stroke volume. We have tested hemodynamics in our laboratory by using a cost-effective, powerful microcomputer-based portable noninvasive technique, which makes possible the ensemble averaging of impedance cardiographic waveforms. in conjunction with physiologic maneuvers, the technique has been implemented at our institution and has provided helpful information in our experience in evaluating volume overload, hypertension, hypotension, shock, and heart failure. It is hoped that this noninvasive, relatively cost-effective approach will be more widely appreciated in the future, given the economic realities of medicine today. (Am Heart J 1988; 116:657.)

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Unless an affordable, atraumatic, replicable method of measuring hemodynamics exists, it appears that a discussion about the relative merits of hemodynamic approaches to hypertension is relatively moot. The difficulty with invasive studies in measuring cardiac output and vascular resistance in hypertensive patients is that such measurements are necessarily infrequent and few, constituting relatively isolated observations in a slowly changing and progressive disease that requires serial observations over many years. This slow evolution of hypertension represents a major dilemma for realistically advocating a hemodynamic approach to hypertension evaluation and management.

Within this context I believe that impedance cardiography, coupled with blood pressure monitoring and physiologic maneuvers, represents a reasonable approach to the need for an affordable, noninvasive means of reflecting cardiac output and hemo dynamics in hypertensive patients.

Although impedance cardiography was introduced as early as 1930, it was only studied systematically in the late 1960s by the National Aeronautics Space Administration to meet the requirements for noninvasive hemodynamic monitoring during the Apollo space flight. Much remains to be understood about the physics of bioelectrical impedance and the behavior of electrical resistivity in tissue, including the waveform and its accurate quantification. Nevertheless, the most appealing aspects of impedance cardiography and its clinical use today are its low cost and nonreliance on highly trained and skilled technicians to perform the tests compared with echocardiography or radioactivity-based techniques. In my opinion the accuracy of impedance cardiography is acceptable at the clinical level, especially considering that the measurement of stroke volume is an imperfect effort at best, even by invasive techniques. [1-2]

PHYSICAL THEORY OF IMPEDANCE CARDIOGRAPHY

The resistance to alternating current flow is known as electrical impedance. If current remains constant, this resistance or impedance is inversely proportional to voltage. The properties of a conductor are related to the inherent resistance of the conducting medium, the length of the conduit, and its mean cross-sectional area. Modern impedance cardiographic instruments inject a low-energy, high frequency—alternating electrical current through the thorax. Another pair of detecting electrodes, always located inside the current path, displays the impedance changes. The frequency of the delivered current is between 20 and 200 kHz, so low in energy that it cannot be sensed by the patient and is totally safe. The average basal electrical impedance inherent in

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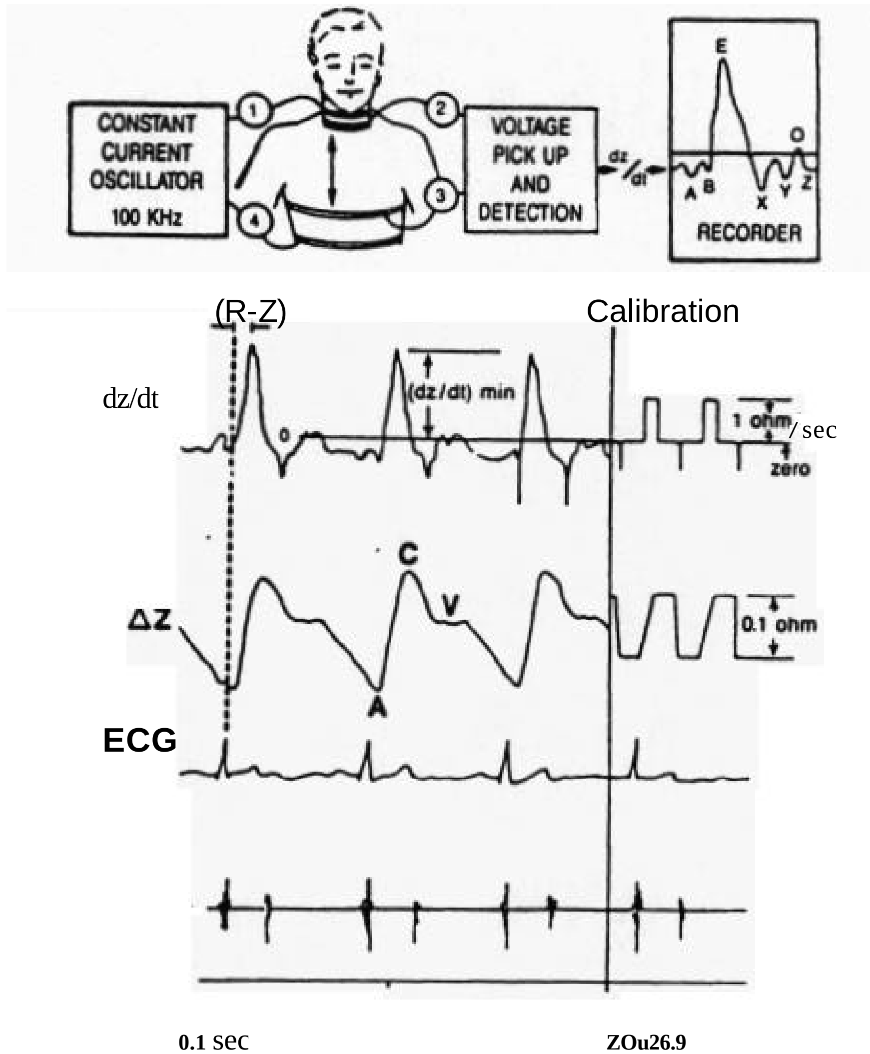


Fig. 1. Morphology and time signatures of impedance cardiography waveform

the thorax is expressed as Zo. Several conductive components residing within the thorax contribute to Zo. Air is a poor conductor; thus during inspiration Zo will rise. Since blood and fluid are good conductors of electricity, accumulation of fluid within the thorax will result in a decrease in Zo.[3-10] The technique has been used for thoracic fluid monitoring, and to the extent that chest fluid equilibrates with total body water, [11-13] may be used as a reflector of extra-cellular fluid volume relative to lean body mass. Zo may be corrected for the length between the measuring electrodes, and there is a relationship between Zo corrected for length and extracellular fluid volume to lean body mass ratio.[14] In addition, during inspiration and expiration there are major shifts in thoracic resistivity as a function of changes in the resistive elements of the lungs.

Blood is the most electrically conductive substance in the tissue under study. Pulsatile flow results in blood volume changes that are phasic; thus electrical impedance will shift phasically as a function of the volumetric changes of blood in the arteries within the segment under study. The change in electrical impedance as fluid pulsates within the measured segment is displayed as ΔZ, a small wave that rides the crest of the respiratory shift. This ΔZ waveform from the impedance cardiogram is similar in morphology and timing to typical waveforms recorded by flow or pressure transducers from the great vessels. The pulsatile signal, or ΔZ, is frequently differentiated in clinical impedance cardiography to yield a dZ/dt signal, which is similar in morphology and tuning to that seen using aortic flow probes.

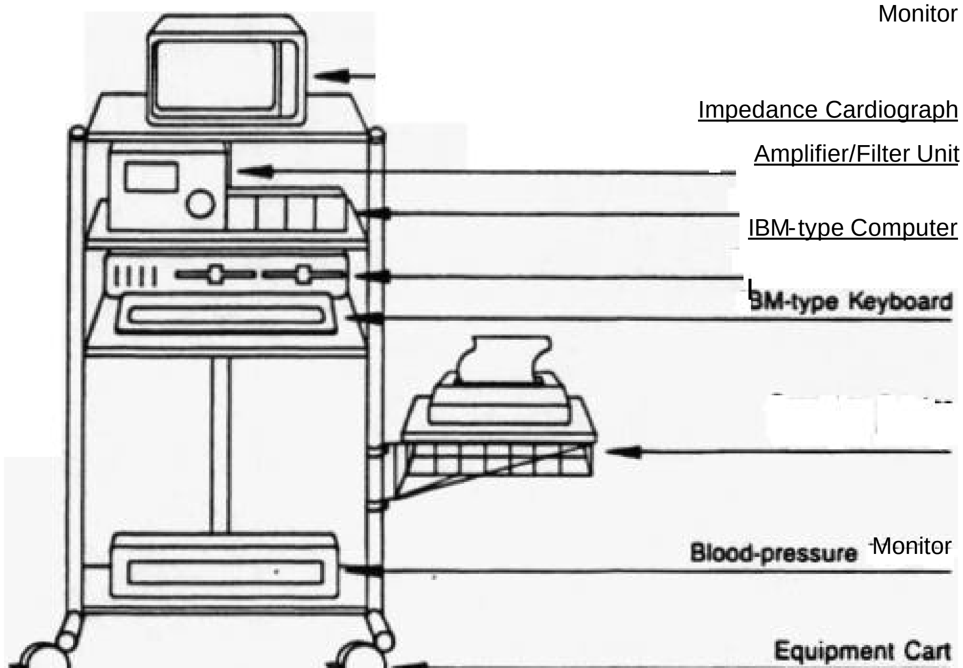


Fig. 2. Cardiac performance laboratory.

GENESIS OF THE IMPEDANCE CARDIOGRAPHIC WAVEFORM

To clarify one possible point of confusion, it should be realized that "peak" flow values actually occur at minimum impedance values. However, the dZ/dt signal is electrically inverted display purposes so that the greatest flow is at the peak rather than nadir of the signal. The typical ΔZ wave (Fig. 1) consists of the following three segments'6: (1) The A wave corresponds to the P wave of the ECG and represents the atrial contribution to ventricular filling; (2) the C wave corresponds to left ventricular ejection, which terminates at the beginning of the incisura or dicrotic notch; and (3) the V wave is initiated approximately with the onset of the second heart sound as an incisura and represents protodiastole.

In clinical impedance cardiography, the first derivative of ΔZ is frequently used, and several waveform components of this dZ/dt signal have time signatures and morphologic characteristics typical for certain cardiac events. The A wave is a negative-going signal related to atrial contraction and follows the P wave of the ECG. The B wave occurs at the time of peak oscillations of the first heart sound. The crossover point, at 0 O/sec of the dZ/dt waveform, signals the onset of left ventricular ejection with its peak derivative, the E point, which coincides with peak aortic flow rate. The X wave signals aortic closure, and the Y wave signals pulmonary valve closure. The O wave signals mitral valve opening, and the Z wave or late O wave is synchronous with the third heart sound.

Experimental animal studies indicate that most of the ΔZ waveform originates from aortic flow; an isolated aortic occlusion causes a marked decrease in dZ/dt amplitude and a distortion of morphology. Conversely, isolated pulmonic occlusion will cause some decrease in dZ/dt height, but the morphology of the waveform stays the same. It is estimated that approximately 60% of the signal originates from left heart flow and 40% from right heart flow. Because of the characteristic time signatures of the dZ/dt waveform, the impedance cardiogram may be used to obtain systolic time intervals. The period from onset of the Q wave to the crossover point of dZ/dt represents preejection. The period from the crossover point to X point on the dZ/dt waveform, the second heart sound on the phonocardiogram, is the left ventricular ejection time.[16-19]

In addition, other indexes of contractility may be obtained by impedance cardiography. The Heather index, which is defined as the ratio of dZ/dt height (ohms per second) to QZ interval (second), is an expression that increases as contractility increases.[20] The impedance cardiographic morphology or magnitude is influenced by many changes in hemodynamic factors, including myocardial kinetics, valve function, tissue conductivity, cardiac rhythm, left ventricular filling volumes, thoracic, pleural, and pericardial fluids, aortic compliance, thoracic habitus, and afterload. Large O or Z waves will be seen in the presence of significant mitral regurgitation[21] whereas marked deepening of the X trough will occur in aortic regurgitation.[22,23] O waves will also be seen during attacks of angina,[24] and a large O wave will appear in congestive heart failure. In some forms of heart disease, the hemodynamics will be so deranged that the O wave will be the most prominent wave observed, exceeding the ejection wave in height. Large O waves tend to be associated with a poor long-term prognoais[25,26]

An A wave on the dZ/dt occurs at the time of atrial contribution to ventricular filling.[27,28] The dZ/dt maximum of the A wave correlates with left atrial ejection fraction,[29] and a deep, large A wave is found in the noncompliant left ventricle.[30]

QUANTITATIVE METHODS IN IMPEDANCE CARDIOGRAPHY

The original model for calculating stroke volume was described by Nyboer[31] in 1939. It assumed a conducting segment of specified length in centimeters, known as L, between sensing electrodes at the thoracic inlet and outlet. Rho is the specific resistivity of blood. The basal impedance was expressed as ΔZo, with a pulsatile impedance component known as Z. The impedance change, ΔZ, was considered equal to the value of an extrapolated systolic downstroke at the dicrotic notch to compensate for venous runoff. The Nyboer equation is:

Stroke volume — Rho * (L7/Zo[22]) * ΔZ.

Because the ΔZ waveform showed considerable respiratory variation, Kubicek[32] subsequently differentiated the signal and using dZ/dt developed the following equation: Stroke volume = Rho * (L^2/ Zo^2) * dZ/dt *T, where dZ/dt is the first derivative of ΔZ. The equivalent to left ventricular ejection time in seconds, and Rho is the specific resistivity of blood. The relative importance and accuracy of determining blood vs thoracic resistivity remains controversial.[33-35]

Sraznek[31] has proposed that the term Rho * L^2/ Zo^2 be replaced with the term V/Zo where V is the volume of thoracic electrically—participating tissue and is equal to L^3/4*.25.* This is based on the supposition that the thorax is a truncated cone and blood resistivity is a trivial factor in total resistivity for the equation.

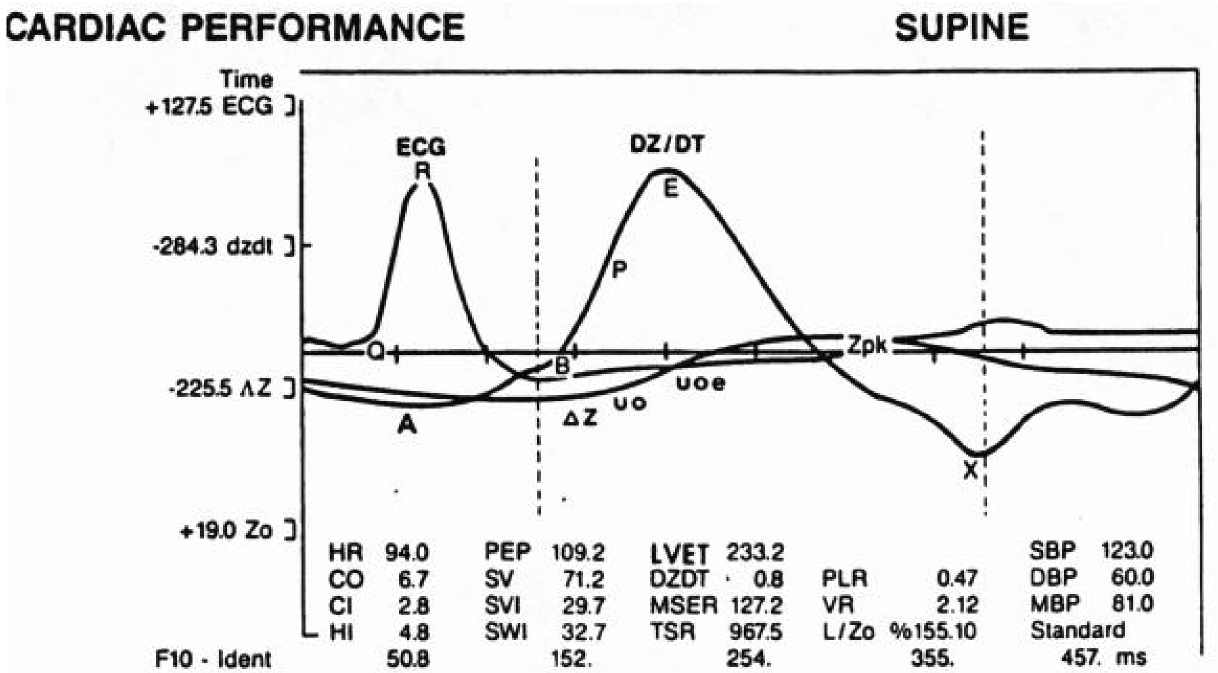
Body habitus appears to play a role in that resistivity is assigned a larger value for those who are obese compared with those with higher muscle-to-fat ratios.[36] In general, women and infants show higher basal impedance or Zo than do men[37] because of a higher relative fat content of body composition.

Measures of stroke volume by impedance cardiography correlate well in most studies with those obtained by invasive techniques, although shunt and regurgitant lesions may create errors since the technique reflects aortic flow. Nevertheless, in the absence of valvular disease, correlation coefficients between impedance cardiography and invasive methods are quite good. Our studies yielded Pearson correlation coefficients of 0.7 to 0.8 when the technique is compared with thermal dilution—derived cardiac output determinations in the cardiac catheterization laboratory.[36] (The reader is referred to several articles reviewing validation of impedance cardiography against other measurement techniques.[38-47])

CLINICAL APPLICATIONS OF IMPEDANCE CARDIOGRAPHY

The hemodynamic approach to hypertension evaluation and management is based on the simple equation that blood pressure equals cardiac output times total systemic resistance. Thus the interest in using impedance cardiography together with blood pressure determination has come about as a means of reflecting hemodynamics. Suitable technology for such a system is represented by the Cardiac Performance Laboratory. The portable impedance cardiographic and blood pressure monitoring device uses a specially configured International Business Machine personal computer and consists of a monitor, a keyboard, a graphics printer, and an amplifier-filter unit (Fig. 2). The device ensemble averages as many as 250 beats of impedance cardiographic waveforms and determines appropriate points on the averaged waveform for the necessary stroke volume calculations.

The computer monitor displays four channels of continuous real-time waveforms that may be frozen and copied on the computer printer at any time. After input of appropriate demographic data and calibration of each signal via hardware and software adjustments of the A-to-D converter, data is simultaneously displayed and acquired in real time for ensemble averaging, which follows an appropriate period of data collection. The ECG and the dZ/dt and ΔZ averaged waveforms are displayed with each printout, and the digitized data may be saved on disk for future review and analysis. Because oscillometric



HR — Heart Rate
CO — Cardiac Output
CI — Cardiac index
HI a Heather Index
PEP — Preejection Periods
SV — Stroke Volume
SVI — Stroke Volume Index
SWI — Stroke Work Index

LVET — Left Ventricular Ejection Time
DZIDT — Height of the DZIDT in ohms per second
MSEA a Mean Systolic Ejection Rate
TSR — Total Systemic Resistance
PLR PEP to LVET Ratio

VR — Vascular Rigidity (pulse pressure/stroke volume index)
L/Zo — Length to Zo Ratio (reflection of extra cellular fluid volume)
SBP = Systolic Blood Pressure
DBP— Diastolic Blood Pressure
MBP- Mean Blood Pressure

Fig. 3. Cardiac performance printout.

blood pressure determinations are automatically taken and entered during waveform collection, a complete set of pressure- and flow-derived functions are calculated and displayed by the computer. The various points used for making the calculations are also labeled on the waveforms displayed so that the operator can actually see the analog signals with wave-determined data points used for the calculations (Fig. 3), and judge their correctness. These calculation points may be moved at the operator's discretion when the computer-determined point needs to be changed. Because this unit is mounted on an equipment cart, it can be taken to the bedside, the cardiac catheterization laboratory, the dialysis unit, or any inpatient or outpatient environment.

Validation of cardiac output determinations by this ensemble-averaged technique has been simultaneously compared with thermal dilution—determined cardiac outputs, which yielded correlations similar to our previous experience.[36,49] In addition, the device can be used in conjunction with physiologic maneuvers to evaluate the hemodynamics of a variety of conditions, including hypertension. Because the technique lends itself well to studying hemodynamics in a variety of postures and positions, we can assess preload, afterload, and ventricular function through postural changes in a physiologically based manner. Although impedance cardiography cannot directly measure filling pressures, it is possible to reduce preload and assess its effect on stroke volume through postural changes such as moving from the supine to the standing position. End-diastolic volume will always be higher in the supine than in the standing position. Therefore by comparing cardiac output and stroke volume in supine and upright positions, one can decide whether the patient's ventricular function is operating on the ascending, the plateau, or the descending limb of the Starling curve (Fig. 4). When the patient assumes the upright position, the normal force of gravity results in pooling of blood volume in the legs, a marked drop in end-diastolic volume, and reflex veno and arteriolar constriction, with a compensatory

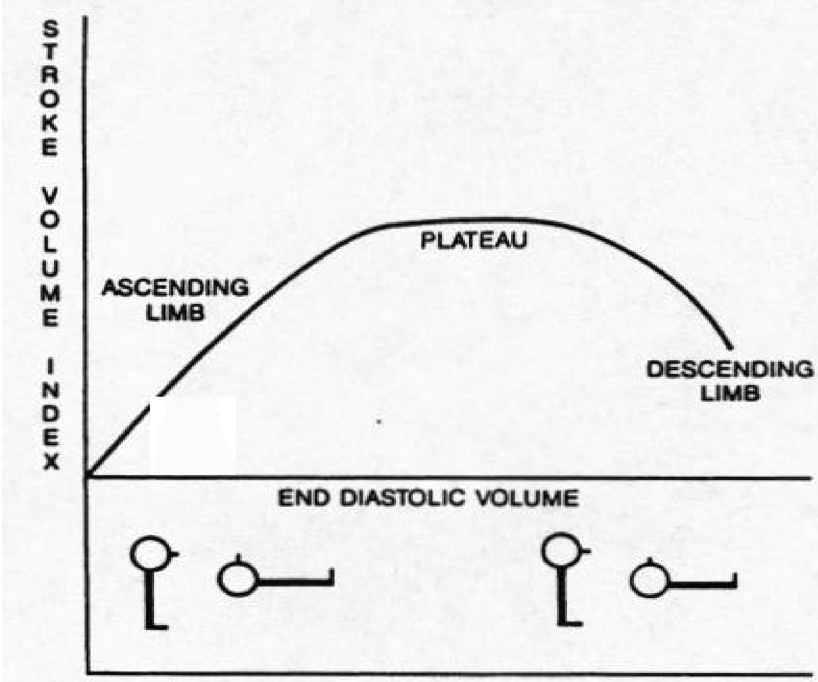


Fig. 4. Postural effects on stroke volume to assess ventricular function.

increase in heart rate. When a normal patient assumes the upright position, the following changes occur (percentages are approximates): a 35% increase in heart rate, a 15 mm Hg rise in diastolic blood pressure, a 40% increase in total peripheral resistance, a 40% drop in stroke volume, and an 18% to 20% drop in cardiac output.[5°] Conversely, patients with decompensated heart failure may actually manifest improved stroke volume in the upright position, whereas those with compensated heart failure who are on the plateau phase of the curve will likely show no difference in stroke volume, whether in the standing or supine position.[5~2]

When computer-averaged impedance cardiography data and blood pressure measurement are combined with a few simple physiologic maneuvers, it is possible to accumulate most of the data necessary to make informed decisions in managing volume overload, hypertension, hypotension, shock, and heart failure in an inexpensive, replicable, noninvasive manner.

For example, in the management of hypotension if arising from the supine position results in profound drops in stroke volume, whether in orthostatic hypotension or shock, one must assume that hypovolemia is present. Conversely, in the hypotensive patient in heart failure, the stroke volume actually rises on standing, which indicates heart failure or cardiogenic shock. Total systemic resistance will tend to be high in patients in both hypovolemic and cardiogenic shock with a low stroke volume. Conversely, in gram-negative septicemia or shock, total systemic resistance is low and cardiac output increases, which represents the body's response to intense arteriolar dilation. The technique owes much of its value to its ability to reflect changes in stroke volume during dynamic maneuvers. At our institution, the test is frequently used in patients treated for heart failure because serial studies may be performed to assess the degree of change or improvement in hemodynamics after preload- or afterload-directed drug therapy is implemented. Patients with renal failure or a renal transplant represent another example of complex volume—resistance relationships that are much more rationally managed when the hemodynamics are known, especially without costly or invasive methods,[53-58] By defining the homeostatic defect in patients with a hemodynamic problem, one might better direct therapy to the mechanism at fault.

To place this approach to physiologic measurement in perspective, we might compare it to the two noninvasive techniques that are relatively ubiquitous in the practice of cardiology today, the ECG and the echocardiogram. The cost of a computerized impedance study is intermediate between the two. The level of skill and training a technician needs to operate this computerized system to record impedance cardiogram is similar to that required for recording a 12-lead ECG.

CONCLUSION

Computer-based ensemble averaging of the impedance cardiographic signal, together with automated blood pressure monitoring, represents a cost effective, noninvasive means of assessing hemodynamics in hypertension as well as in many other hemodynamic-related disorders. Although the bioelectrical physics of tissue impedance remains a complex and incompletely understood issue, it appears to have empiric clinical merit. Like electrocardiography, impedance cardiographic waveforms are easy to record. However, the clinical relevancy of various morphologies may be difficult to understand unless incorporated within its clinical context and a thorough understanding of cardiovascular physiology. Impedance cardiography can reflect the timing of cardiac events, shifts in tissue fluid accumulation, indexes of cardiac contractility, flow velocities, ventricular function, and volumetric determination of the central circulation—all from the same relatively inexpensive, noninvasive technology. Although both basic and clinical research remains to be done to further define the biophysical factors influencing the accuracy of impedance cardiography, it is clear that considering today's economic pressures on medicine, we can no longer advocate the mass implementation of ever more expensive technology as a routine approach to evaluating clinical hemodynamic problems. It is hoped that in the future ensemble-averaged impedance cardiography, mated with blood pressure measurement and physiologic maneuvers, can be more seriously regarded as an attractive and empirically useful alternative for obtaining hemodynamic data in patients with cardiovascular disease.

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