ELECTROCARDIOGRAPHY

Electrocardiography is the study of the heart's electrical activity recorded from the surface of the body. Such recordings represent a total or integrated view of all of the electrically excitable cells in the heart. A sensitive medical recording device, called an electrocardiograph, is attached to the body with special electrodes and records the voltage changes on chart paper. This voltage versus time recording is the electrocardiogram. Both the device and its graphical output are abbreviated by the familiar acronym "ECG" and, depending on its contextual use, one could be referring to either one. Because much of the original work in this field was performed by Willem Einthoven in Holland, the abbreviation of "EKG" was based on the Dutch root word "kardio" and is interchangeably used with ECG.

The ECG provides the physician with a significant series of waves from which one can measure the rate, rhythm, and many aspects of the health of the various cardiac muscle tissues that comprise the heart. The actual recording devices have kept pace with advances in modern technology, so that today's recording devices use integrated electronics and embedded microprocessors to record, analyze, and store the signals generated by the heart. In addition, a wide variety of medical devices rely on an ECG signal, in part, to perform their primary function. Examples of these devices are treadmill systems where the heart is monitored under exercise workload conditions; cardiac pacemakers that monitor the heart rhythm from internally implanted electrodes to determine if it is necessary to stimulate the heart because of loss of function of the heart's natural pacemaker; and sophisticated imaging systems that require synchronization with the cardiac cycle to minimize the effects of cardiac motion. Thus, the ECG is still evolving as a tool for studying the heart even though it is perhaps the oldest test instrument in medicine.

Physiological Basis

When two wires are placed anywhere on the body surface and then attached to the inputs of a bioelectric amplifier, it is possible to record the voltage generated by the heart. There are standard positions for placing the recording electrodes on the body, but generally the potential difference measured between any two recording sites is the summation of electrical signals generated by billions of cardiac cells. The adult heart is a bit larger than a fist, and the sequence of its electrical activation is directly related to the contractional sequence of the various heart chambers. It is important to note that the electrical signals are the triggering event for the mechanical motion of the heart and that these electrical events precede the heart contraction. The electromechanical coupling is a significant phenomenon for the overall function of a healthy heart, but it is possible to have an electrically normal heart while the mechanical function could be significantly impaired and vice versa. This article focuses on the electrical activity of the normal heart, for which it is important to understand some fundamental aspects of the heart's anatomy and physiology (1).

There are four chambers in the heart. The two upper chambers are the atria, and the two lower chambers are the ventricles. Another way to divide the heart is into the right and left side with the result that the four chambers are the right atrium, right ventricle, left atrium, and left ventricle. Figure 1 is a schematic representation of the four chambered heart with its physical connections to the veins which deliver blood into the heart and the arteries, the vessels that carry blood away from the heart. One could begin anywhere in the

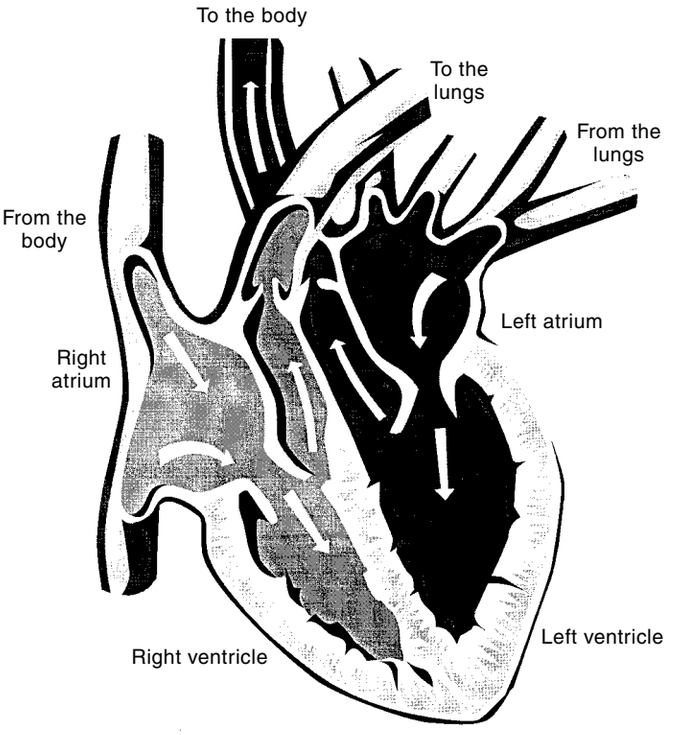

Figure 1. A cutaway diagram of the heart showing the major chambers and vessels. The blood flow into and out of the heart is indicated by the arrows.

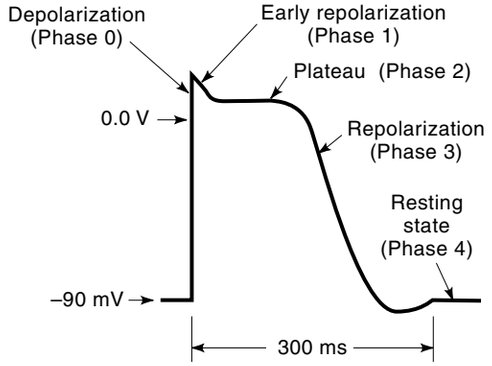

Figure 2. A cardiac action potential recorded with a microelectrode inside of a single cardiac cell. There are five phases describing the three electrical states of each cell.

circulatory system to describe the sequence of events that make up the cardiac cycle, but the most common starting point is the right atrium. This approach takes the view that there are two pumps in series, each with a low-pressure collecting chamber, the atria, and each with a high-pressure pumping chamber, the ventricles. In fact the atria contract sequentially whereas the ventricles contract together. This parallel view of the cardiac cycle is examined when discussing electrical activation of the heart.

The right atrium collects blood from all of the veins in the body except those of the lung. When the right atrium is triggered to contract, it forces blood into the right ventricle. As the right ventricle fills, it contracts and forces blood to the lungs where the blood exchanges the excess carbon dioxide, an end product of metabolism, for more oxygen, a necessary metabolic component. The pulmonary veins return this oxygen-enriched blood to the left atrium that in turn empties into the left ventricle. The left ventricle is a high-pressure pump that forces blood to all of the body organs and tissues except of course the lungs. This is done through the arterial blood vessels which evolve to microscopic tubes called capillaries which then reform into larger vessels forming the venous return system to the heart.

Before discussing the electrical activation sequence of the heart, it is necessary to understand some of the basic concepts of cellular electrophysiology. A complicated cellular process of ion flow across the cell membrane occurs with each heartbeat. It is possible to monitor these currents and their resulting transmembrane potentials when cardiac tissues are studied in vitro, or outside the body, and placed in a warmed, nourishing solution. Such studies form the basis of almost all current knowledge of cardiac electricity, but a detailed discussion of this work is well beyond the scope of this report. The article BIOELECTRIC PHENOMENA in this encyclopedia covers this material. Figure 2 is a recording of a transmembranous action potential from a microelectrode that can actually be impaled into a single cardiac cell. It shows how the voltage from the inside of a cardiac cell varies with respect to the outside of the cell and how it varies during the time of cardiac activation. Cardiac cells have three electrical states that comprise the cardiac cycle, the resting state, depolarization, and repolarization. Scientists who study the cardiac action potential refer to the resting state as Phase 4. The electrical states and phases of the action potential are labeled along the waveform at points representing their respective occurrence. Hence the resting state and Phase 4 are synonymous. The resting state is the time between beats and, depending on the heart rate, is the most common state of the heart. For example, cardiac action potentials range in duration from 200 ms to 350 ms, so if the heart rate is 60 beats per minute (1 bps), then the resting state accounts for 65% to 80% of an individual cell's electrical activity. During the resting state the inside of the cell has a negative potential of about −90 mV with respect to the outside of the cell, which is assumed to be at 0.0 V.

As an aside to this discussion it is interesting to note that this potential difference occurs across a very small distance, the thickness of the cell membrane. This distance is on the order of 1 nm. This results in a field strength of (90 mV)/(1 nm) = 90 MV/m. This is a very large electric field considering that air conducts electricity, for example, static spark or lighting, at field strengths of about 300 kV/m. Maintenance of this large electric field across the cell membrane is most likely due to the lipid (fat) bilayer which forms the bulk of the cell membrane.

Activation of a cardiac cell occurs when ionic currents flow from neighboring cells or from a nonphysiological stimulus such as a mechanical impulse or a pacemaker battery pulse. The membrane potential of a single cell can shift very rapidly from −90 mV to 20 mV (a total swing of 110 mV) in approximately 1 ms. This rapid change is called depolarization or Phase 0 of the cardiac action potential. This rapid change in membrane potential causes neighboring cells to depolarize, and thus the cardiac impulse spreads throughout the myocardium from cell to cell. Although the stimulus for initiating depolarization may vary, a key factor required for depolarization is changing the resting membrane potential to a specific threshold voltage. Once the membranous potential reaches this threshold, the individual cellular dynamics take over, and now all of the following phases are independent of the original stimulus. The speed at which cell-to-cell propagation spreads across and through heart tissues depends on the particular cell type and is discussed later.

Immediately following the depolarization state is the return to the resting state, called the repolarization state. This process has three well-defined phases. Phase 1 is an early, rapid, low-amplitude decline in potential. Phase 2 is a relatively isopotential period called the plateau, and Phase 3 is the return to the resting state.

All of the cellular events previously described imply that each cell acts as a tiny current source with a resulting potential field. The current spreads throughout the body because all of the tissues enclosed within the skin form what is called a volume conductor and, for the most part, these tissues are good conductors of electricity. Thus, what is measured on the body surface or from within the body for that matter is a summed view of all of the individual potential fields generated by each individual cell.

The heart's primary function is to pump blood. This mechanical process is triggered by an electrical signal generated by each cell once it is stimulated. Cardiac muscle cells are bricklike in appearance. They have a base which is about 10 μm on each side and are about 100 μm long. Rather than viewing activation and contraction cell-by-cell, it is often simpler and more convenient at times to consider these electrical and mechanical processes on a more macro or tissue level. Hence one could refer to atrial activation or ventricular con-

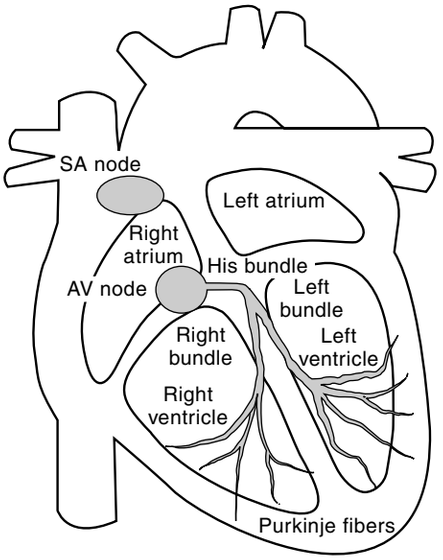

Figure 3. A schematic diagram of the cardiac conduction system. The major components of the conduction system are shaded and shown in their approximate anatomic position, but they are not to scale.

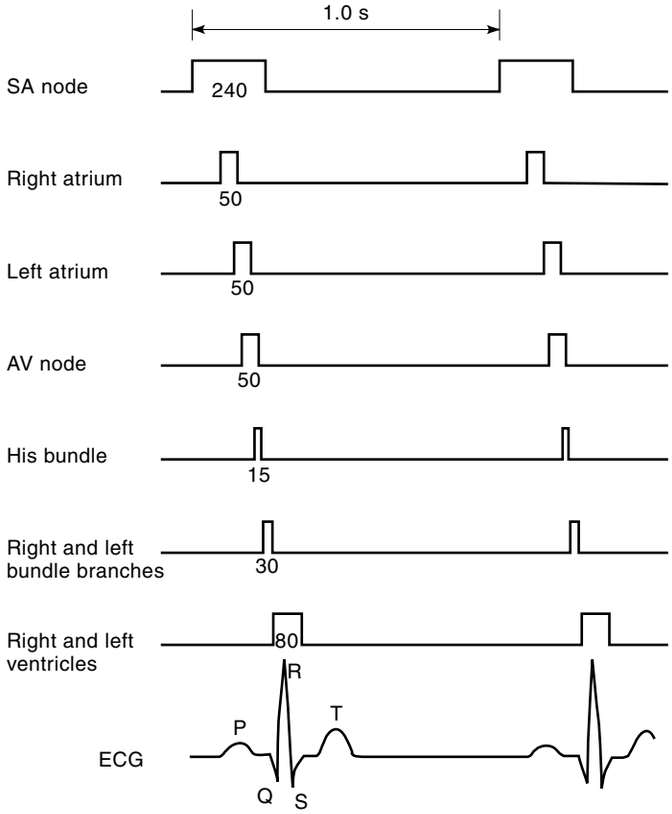

Figure 4. A timing diagram showing the activation times for each of the major muscle chambers and the conduction system. Note that the durations of the positive pulses (labeled in milliseconds under each pulse) indicate the time during which cells in each structure undergo depolarization. The bottom trace is a stylized ECG showing the correspondence of atrial and ventricular depolarization with the surface manifestations of the P-wave and QRS-complex, respectively.

traction. This integrated view of the heart's mechanical function is called a syncytium (1).

Now let us consider the heart on a more anatomical basis and describe the actual sequence of activation and how this relates to generating the measured waves of the ECG. Figure 3 is a cross-sectional sketch of the heart. All four chambers are respectively labeled. A network of structures has been added that comprise the heart's specialized conduction system. Consider Figs. 3 and 4 together to understand the sequence of cardiac activation better. Figure 4 is a timing diagram that indicates the length of time during which each particular structure has cells undergoing phase zero depolarization. The bottom trace in Fig. 4 is a stylized ECG recording whose component waves are labeled allowing one to compare the surface waves with the internal cardiac sources. Note that not all of the internal structures are observed on the standard ECG.

The electrical activation of the heart begins with automatic depolarization of an irregular mass of cells in the upper portion of the right atrium called the sinoatrial (SA) node. Once these cells end their repolarization, there is a gradual incline of Phase 4 toward the threshold voltage. Thus these cells do not require an initial excitative current from any other cell, and they comprise the heart's natural pacemaker. The right atrial muscle cells respond to these neighboring and depolarizing SA nodal cells by initiating their own depolarizing currents. Notice that there is considerable timing overlap between the SA node and the right atrium because the SA node does not have a single point of interconnection with the right atrium. The speed at which the cardiac impulse travels is relatively slow in the SA node, on the order of 0.05 m/s.

Conduction velocity plays an important role in cardiac activation. It is easily measured by placing two electrodes with known spacing on the heart. The occurrence of depolarizing voltages is easily identified at each site. Together with the time measured between these two events, the conduction velocity is determined. Conduction velocity is the ratio of the time between the two events divided by the distance between the two recording sites. Table 1 lists conduction velocities for the various cardiac tissues. Note also that physiologists refer to conduction velocity as the measure of speed of activation through electrically active tissues. In this sense conduction

Table 1. Conduction Velocity of Cardiac Tissues[a]

SA Node	Atrial Muscle	AV Node	His Bundle	Bundle Branches	Purkinje Fibers	Ventricular Muscle
0.05	0.5	0.05	1.0	1.0–2.0	4.0	0.5

[a] All values are given in m/s.

velocity is a scalar measurement. In addition the bricklike shape of the cardiac cells and their regular pattern of end-to-end construction with staggered rows on top of each other also affect the speed of activation. In the ventricles, for example, the speed of conduction is three to four times faster in the direction of the long axis of the cells than across their short axis. The reason for this is based on the number of interconnections which exist at cell ends rather than at the sides of the cells. This directionally dependent property is called anisotropy. Hence, if one artificially stimulates the ventricles at a single site, the patterns of activation away from this site are generally in concentric ellipses rather than concentric circles.

Atrial activation spreads from right to left and from top to bottom. The timing diagram shows that the duration of SA nodal activation is about 240 ms. Some of the right atrial muscle begins activation during the period of SA nodal activation. The right atrium begins activation about 50 ms prior to the left atrium. Total atrial activation takes about 100 ms with a conduction velocity of about 0.5 m/s. As seen in the bottom trace of Fig. 4, the ECG registers the P-wave during atrial depolarization. When the activation wave front reaches the lower floor of the right atrium, another nodal structure called the atrioventricular (AV) node is activated. This is also a slowly conducting structure (0.05 m/s) and is considered the initial conduit of activation from the atria to the ventricles. Although the atria and ventricles border each other, the only electrical connection between them is through the AV node. The relatively slow conduction through the AV node allows time for the atria to contract and fill the ventricles. It takes about 50 ms to 70 ms for activation to traverse the AV node as indicated by the timing diagram. The AV node connects to a small fiber, about 1.0 mm in diameter and about 10.0 mm to 15.0 mm long, which actually leaves the right atrium and penetrates to the top of the ventricles. This fiber is called the common bundle or the bundle of His, named after Wilhelm His, a German physiologist, and his early work on conduction system anatomy, although he was not the first person to actually identify this structure. Conduction velocity accelerates through the His bundle to about 1.0 m/s, and it has a total activation time of about 10 ms to 15 ms. At this point the electrical impulse is traveling atop the wall or septum that separates the left and right ventricles. The His bundle and its early ramifications are not electrically connected to the ventricular muscle tissue. These ramifications separate into bundles on the right and left sides of the septum and are appropriately called the right and left bundle branches. Conduction velocity accelerates to about 2.0 m/s and lasts about 30 ms to 35 ms in these bundles. A further arborization of this conduction system tissue occurs over the inner surface of the ventricles, called the endocardium. The Purkinje fibers are the final stage of the conduction system and form a fine fiber mesh covering over some parts of the endocardium. This widespread, thin layer of tissue is where the conduction system and ventricular muscle tissue finally form electrical interconnections. A large part of the endocardium is initially activated by the Purkinje network. This is followed by conduction from muscle cell to muscle cell which results in total electrical activation of the ventricles. The conduction velocity through the ventricular muscle is about 0.5 m/s, and it takes about 80 ms to 100 ms to depolarize the ventricular myocardium. The timing diagram shows that the two ventricles are activated at about the same time and have roughly the same duration of activation.

A sample ECG tracing is shown at the bottom of the timing diagram. The first rounded wave, called the P-wave, is the body surface's manifestation of atrial depolarization. Toward the end of the P-wave the AV node and His–Purkinje system depolarize. The standard ECG does not show any evidence that these structures depolarize because they produce very small signals compared with the large muscle masses of the atria and the ventricles. A specialized branch of electrocardiography, called high-resolution electrocardiography (2), uses computer-based enhancement techniques to record these low-level signals from the body surface, which are described later in this article. Once the ventricles are depolarized, the ECG shows a rapidly changing voltage called the QRS-complex. The initial downstroke of this complex is called the Q-wave, and the initial upstroke is called the R-wave. The final downstroke is called an S-wave. The nature of the QRS-complex depends highly on the specific lead and any underlying cardiac abnormalities. Thus, there are times when the ventricular depolarization wave may have only a R–S-pattern, or may be just a deep Q-wave. Generally, the complex is called a QRS-complex even if the pattern does not strictly follow the Q–R–S-sequence. Following the QRS-complex is a smooth wave called the T-wave. This wave represents the repolarizing currents of the ventricles. Notice that the atria do not produce a similar repolarizing waveform. The atrial repolarization wave temporally overlaps with the QRS-complex and is masked by its higher voltage.

Figure 5 is a larger, stylized ECG recording showing both amplitude and timing scales. In ECG terminology an "interval" is the period from the beginning of one wave to the beginning of the next wave. An example is the PR-interval shown in this figure. A "segment" is the period between two waves as demonstrated by the ST-segment in this figure. Finally this tracing shows the U-wave, a small wave after the T-wave. The U-wave is associated with repolarization, but its actual origin has yet to be definitively determined.

Measurements of the amplitude and duration of each wave and the above-mentioned intervals and segments are used by electrocardiographers to diagnose cardiac pathologies such as ventricular enlargement (hypertrophy), blocks of conduction in the bundle branches, damage to the ventricles due to heart attacks, and abnormalities of rhythm. Such diagnostic interpretation of the ECG is an important skill, and there are a number of texts devoted to the subject (3).

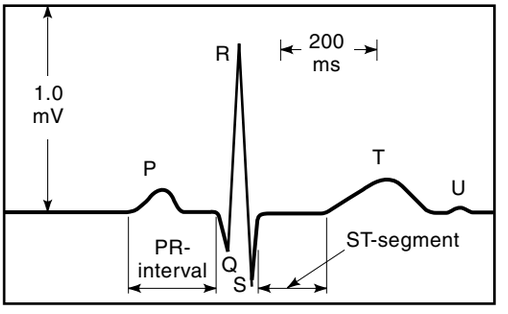

Figure 5. A stylized ECG recording showing the individual waves (P, QRS, T, U) and the approximate time and voltage scales. Both the PR-interval and ST-segment are also labeled.

History

The electrocardiogram was first recorded by Augustus D. Waller, an English physiologist, in 1887. His pet bulldog, Jimmie, was his first subject using a device called the capillary electrometer. The device, crude by today's standards, used a voltage-sensitive column of mercury that reflected a beam of light from its meniscus onto a moving photographic plate. Waller is credited by many to have actually coined the term electrocardiogram. However, Willem Einthoven, a Dutch physiologist, is usually credited with bringing the ECG into clinical practice with a string galvanometer. This device used a thin wire between poles of a magnet. A movement of the wire occurred which was proportional to the current flow. Motion of the wire could be used to scrape a carbon residue off a slowly rotating "smoked" drum. The evolution of ECG recording instrumentation closely followed the developments of electronic technologies, such as the vacuum tube, the transistor, the integrated circuit, and the microprocessor. The use of computers to automate the interpretation of the ECG was a very early application of computers in medicine (4). Since the 1960s the algorithms for a fully automatically interpreted ECG have been developed and optimized to the point where physician overreading, although still a technical requirement, is seldom necessary for normal ECGs. Complex arrhythmias have proven difficult for automated interpretation. It usually requires a highly trained physician to render an accurate reading. Systems in use today often resemble a fully functioning computer system with very specialized software. Figure 6 shows a block diagram of a modern ECG system in which embedded microprocessors are used to digitize and analyze the 12-lead ECG. In addition, microprocessors control the digital input/output systems which include a graphics screen, full keyboard, strip chart recorder, and floppy disk drive.

Theoretical Basis

As ions cross the cell membranes, a tiny current source exists inside the body. The body generally consists of fluids, muscles, fat, and other organ tissues which all act as good conductors of electricity. The body is bounded by a skin interface and hence is referred to as a volume conductor. Theoretically, the potential field established by the flow of ionic currents in the body tissues appears at all points within and on the surface of the body. Because of the large differences in the resistances between the environment and the body tissues these voltage potential fields are not found beyond the body surface. Hence, there is no current flow from the body to free space, or put another way our bodies do not supply electrical energy to the outside world.

For signal frequencies in the range of the ECG (0.05 Hz to 200 Hz) the body is considered purely resistive and does not contain inductive or capacitive elements (5,6). Thus, as the heart depolarizes, events on the body surface, which reflect these cardiac sources, appear instantaneously. There is no time delay between changes in the source and the appearance of these events on the body surface. This instantaneous propagation from the heart to the body surface should not be con-

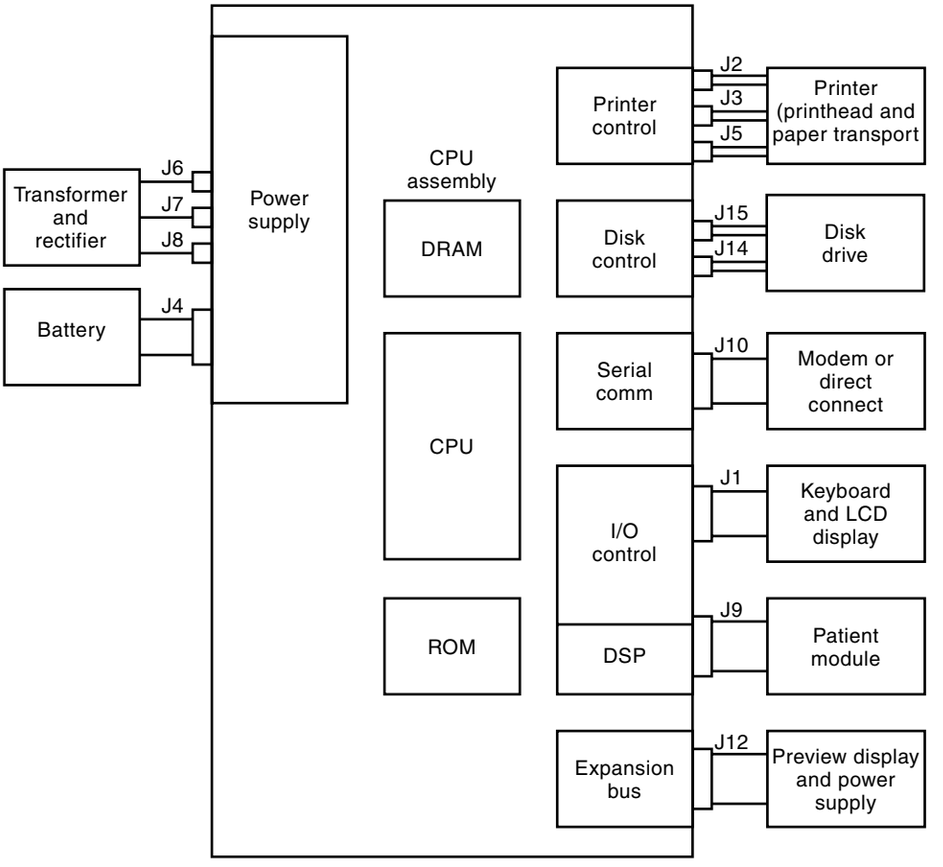

Figure 6. A block diagram of a modern computer-based ECG recorder. Such a device will amplify, digitize, and analyze the ECG and has all the features of a modern embedded microprocessor-based instrument (courtesy of Hewlett-Packard Company, Palo Alto, CA).

fused with the measurable conduction velocities within the heart (see Table 1).

There are literally billions of individual cells in the heart that depolarize during the cardiac cycle. The sequence of depolarization from the various structures within the heart was schematized in Fig. 4. As one proceeds down to the cellular level, there are only groups of cells depolarizing at any given moment. It is possible to represent those groups of cells which are simultaneously depolarizing as an equivalent source. A common representation of the equivalent cardiac source is that of a dipole with a time-varying magnitude, orientation, and position within the body. This representation is a first-order model useful for understanding the generation of the main waves of the ECG recorded from a bipolar lead with electrodes on the body surface.

The equivalent heart source can be expressed as a three-dimensional "heart vector" $\boldsymbol{H}$ in a Cartesian coordinate system. When two electrodes are placed on the body surface, another "lead vector" $\boldsymbol{L}$ is formed by connecting the two electrodes. The voltage recorded between two electrodes from the cardiac source vector is directly proportional to the dot product, $\boldsymbol{H} \cdot \boldsymbol{L} = |\boldsymbol{H}|\ |\boldsymbol{L}|\ \cos(\theta)$. More specifically, the voltage between the two electrodes is given by the component of $\boldsymbol{H}$ in the direction of $\boldsymbol{L}$. This concept is referred to as the lead field theory. To visualize the lead field concept better, we can use the principle of reciprocity. Briefly, this old network theorem states that when a voltage source at one location within a circuit produces a current between two nodes in the circuit, then one can inject the same measured current in these two nodes to duplicate the original voltage at its original location. Figure 7 demonstrates how this can be applied to the ECG. The top panel shows an outline of a torso with a drawing of the heart in its approximate anatomical position. The two dots represent two body surface electrodes with current flowing between them. The lines crossing through the torso and heart represent the flow of current or the "reciprocal lead field." Note that the heart is in the densest region of the current lines. Components of the heart vector lying along the lead axis, i.e., parallel to the lead field, produce the largest relative voltages. In the lower panel the recording electrodes are placed along the right side of the torso. The current lines do not substantially cross through the heart, and hence this lead axis, i.e., parallel to the lead field, would be a poor one for recording the ECG. This concept of reciprocally energizing the recording leads and mentally visualizing the extent of the current lines which pass through the heart is useful in understanding the lead field concept and determining whether two electrodes placed on various body sites will record a large amplitude ECG.

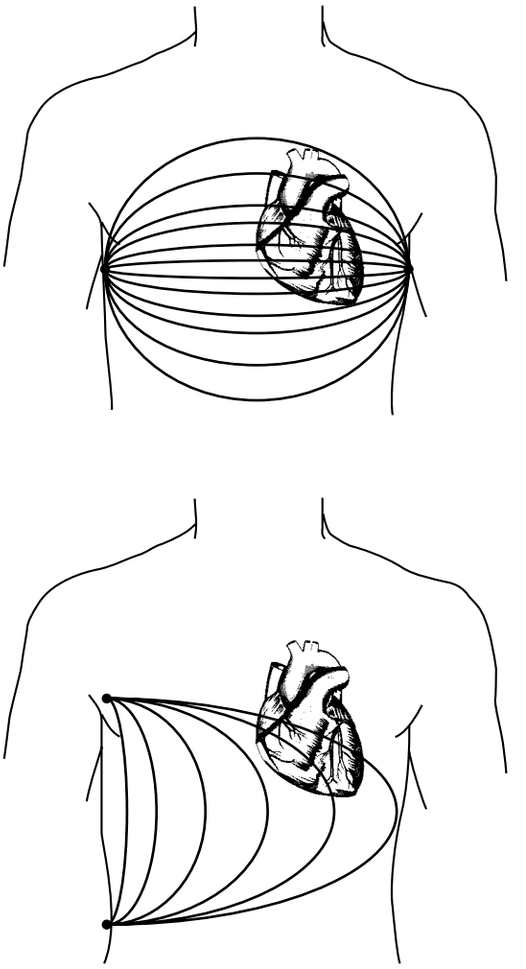

Figure 7. Each panel has an outline of a male torso with the diagram of the heart positioned approximately in its proper place within the chest. The solid dots represent electrodes on the body surface. The lines represent current flow between the electrodes and demonstrate the lead field concept described in the text.

METHODOLOGY

The Recording Technology

The modern ECG system still relies on attaching a set of wires to the skin to couple the potentials on the body surface directly to electronic amplification systems. The electrodes that contact the skin are usually made of a silver/silver chloride (Ag/AgCl) electrode. A conductive electrode jelly or paste acts as the interface between the skin and the metal. This electrode–tissue interface relies on establishing a stable chemical reaction between the ionic charge carriers in the body and the electron charge carrier in the metal electrode (7). The Ag/AgCl electrode is preferred because it is a nonpolarizing electrode through which current freely passes. Generally, this chemical reaction stabilizes in a minute or so and does not interfere or alter the nature of the electrical signal from the heart. There are times, however, when the chemical reaction does not readily equilibrate, for example, when the subject is ambulatory, resulting in a recording susceptible to artifact.

Then the electrode wires are directed to the inputs of a special differential amplifier used to record bioelectric events. These bioelectric amplifiers must meet a number of technical requirements to record the millivolt level ECG signals and to ensure safety when connected to a human being. In general, the bioelectric amplifier has differential inputs with a high input impedance (>100 MΩ) and a bandwidth between 0.05 Hz and 150 Hz. There are two sets of standards which most manufacturers rely on for the ECG. They are published by the American Heart Association (8) and the Association for the Advancement of Medical Instrumentation (9).

Digitizing the ECG signal is relatively straightforward with a standard analog-to-digital converter (ADC). The dynamic range of the ADC is the voltage range over which the analog input voltage is converted to a binary number. Hence the ADC may have an input range of ±1.0 V. Another figure of merit for the ADC is the number of bits in the converted binary digit. Most commercial ECG systems use a 16-bit converter which has a dynamic range of 2^{16} or about 96 dB. This is a very high dynamic range for the typical ECG signal, but it does allow the ECG system to accommodate a widely varying baseline drift which can occur when the tissue–electrode interface is not well established. This artifact, known as baseline wander, can be problematic when interpreting the ECG. There are several digital methods which correct for the baseline wander, and the large dynamic range of the ADC allows these algorithms to operate without the amplified ECG signal reaching the limits of the ±1.0 V ADC range.

Automated Measurements and Analysis

Once the ECG signals are digitized, there are many forms of measurement and analysis that are automatically performed to aid the medical professional in interpreting the clinical information contained in the ECG. The section on applications, covers several of these, but one of the most common functions performed in each application is detecting each beat (10). In this case it is assumed that each beat means every ventricular contraction or every QRS-complex. There are times when atrial and ventricular activity are not synchronous and automated analysis requires detecting both atrial (P-wave) and ventricular activity (QRS-complex). Generally the first step in automated analysis is detecting each QRS-complex. It is the most prominent deflection of the ECG has the largest amplitude (~1.0 mV), and the most rapid change of potential. This rapid change in potential can be detected by taking the derivative (dV/dt) of the ECG and searching for the largest value of the derivative. Of particular concern in this approach is that noise which may contaminate the ECG recording also produces large derivative values, but the noise is not usually larger in overall amplitude than the QRS-complex. Once each possible beat is detected by searching for the largest first derivative, other algorithms can be used to examine the shape of the QRS-complexes and to classify them as normal or abnormal. By measuring intervals, amplitudes, and other wave characteristics, a number of ECG applications can be automated. The following section describes several of these applications where computer based algorithms are used to replace a human operator, and also to create new forms of analysis not amenable to human measurement.

APPLICATIONS

Five ECG applications are presented in this section. In most cases, these instruments are the result of older technology which has matured with the evolution of computer-based instrumentation: 12-lead ECG, monitored ECG, stress ECG, high-resolution ECG, and intracardiac ECG.

The 12-Lead ECG

Einthovin demonstrated the clinical value of the ECG and some of its theoretical underpinnings using the three bipolar limb leads: I, II, and III. These leads are defined as follows:

$$\begin{aligned} \mathrm{I} &= V_{\mathrm{LA}} - V_{\mathrm{RA}} \\ \mathrm{II} &= V_{\mathrm{LL}} - V_{\mathrm{RA}} \\ \mathrm{III} &= V_{\mathrm{LL}} - V_{\mathrm{LA}} \end{aligned}$$

where the terms V_{LA}, V_{RA}, V_{LL} represent the voltages recorded at the left arm, right arm, and left leg, respectively. Note that since each measurement is the difference between two voltages sharing a common reference, the choice of the reference location is arbitrary, and its symbol disappears in the algebra. This is shown in the top panel of Fig. 8. The lines connecting the three limbs define a triangle known as Einthoven's triangle, and it demonstrates Einthoven's law:

$$\mathrm{III} = \mathrm{II} - \mathrm{I}$$

In 1934, Frank N. Wilson (11), an American physician, introduced a concept whereby the signals from the two arms and left leg were averaged by connecting them together with a set of equal resistors, as shown in the lower panel of Fig. 8. This common terminal was used as a reference for other electrodes attached to the body. Thus, new leads were formed using what is now commonly called the Wilson central terminal or WCT. The voltage at the WCT is defined as

$$V_{\mathrm{WCT}} = (V_{\mathrm{LA}} + V_{\mathrm{RA}} + V_{\mathrm{LL}})/3$$

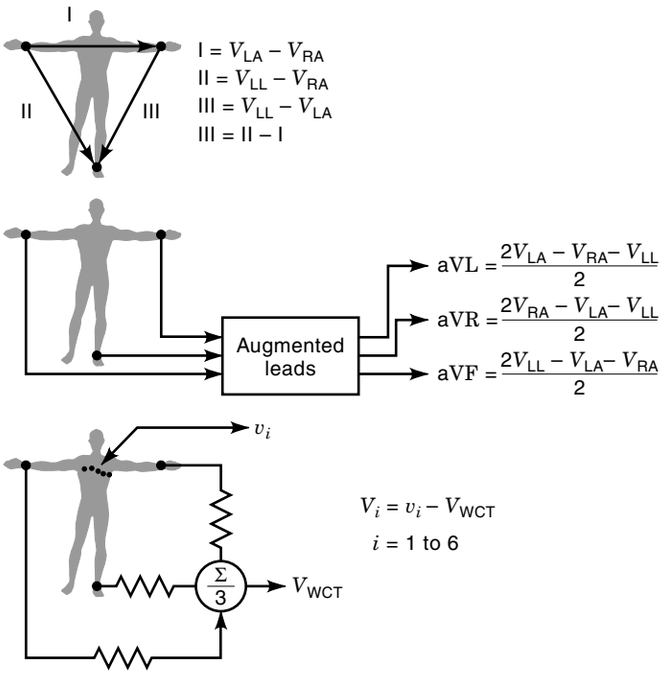

Figure 8. The electrode positions for the 12-lead ECG are demonstrated. The top panel shows the formation of Einthoven's triangle from leads I, II, and III. The voltages on the limbs are from the right arm (V_{RA}), the left arm (V_{LA}), and the left foot (V_{RA}). The middle panel shows the formation of the augmented leads which are linear combinations of the limb leads. The bottom panel shows the Wilson central terminal (WCT) formed by averaging the limb voltages through an equal value set of resistors. The WCT is the reference for the chest leads (V_1, V_2 . . . V_6).

Initially, Wilson proposed a new set of leads abbreviated VR, VL, and VF formed by the pairings of the right arm and the WCT, the left arm and the WCT, and the left foot and the WCT, respectively. Thus the three additional leads are given as follows:

$$\mathrm{VR} = V_{\mathrm{RA}} - V_{\mathrm{WCT}} = V_{\mathrm{RA}} - (V_{\mathrm{LA}} + V_{\mathrm{RA}} + V_{\mathrm{LL}})/3$$
$$\mathrm{VL} = V_{\mathrm{LA}} - V_{\mathrm{WCT}} = V_{\mathrm{LA}} - (V_{\mathrm{LA}} + V_{\mathrm{RA}} + V_{\mathrm{LL}})/3$$
$$\mathrm{VF} = V_{\mathrm{LL}} - V_{\mathrm{WCT}} = V_{\mathrm{LL}} - (V_{\mathrm{LA}} + V_{\mathrm{RA}} + V_{\mathrm{LL}})/3$$

E. Goldberger (12), an American physician, recognized that each term had a duplicated voltage and defined the modified WCT where only the two nonduplicated limb voltages were averaged. These augmented leads were identical in appearance to those originally defined by Wilson but were 33% larger in amplitude. In this era, such an increase in amplitude was significant considering the quality of the amplifiers used to record the ECG. The augmented leads are defined as follows:

$$\mathrm{aVR} = (2V_{\mathrm{RA}} - V_{\mathrm{LA}} + V_{\mathrm{LL}})/2$$
$$\mathrm{aVL} = (2V_{\mathrm{LA}} - V_{\mathrm{RA}} + V_{\mathrm{LL}})/2$$
$$\mathrm{aVF} = (2V_{\mathrm{LL}} - V_{\mathrm{LA}} + V_{\mathrm{RA}})/2$$

Similarly a set of electrodes is attached across the front and left side of the chest and recorded with respect to the WCT. The approximate placement of these chest electrodes shown in the lower panel of Fig. 8 are called V_1 through V_6. The standard 12-lead ECG consists of the following leads: I, II, III, aVR, aVL, aVF, V_1, V_2, V_3, V_4, V_5, V_6. These leads are often formatted in the fashion shown in Fig. 9 where a 2.5s window of each lead is displayed and a 10s period of a single lead for rhythm analysis (bottom trace). Note that the time and voltage scales are 25 mm/s and 1.0 mV/cm, respectively. These scales are relatively low in resolution but have been in use for about 75 years. The 12-lead ECG forms the traditional diagnostic ECG and although a relatively low tech approach is used for display to the physician, the large empirical data base used by physicians will most likely rely on this format for generations to come.

The text in the upper portion of Fig. 9 includes information about the fictitious patient, a set of basic measurements, such as heart rate and interval measurements, and a set of diagnostic statements generated by the interpretive program. The automation of ECG interpretation is usually divided into two phases. In the first phase a set of algorithms is used to obtain a multitude of wave amplitudes, duration, and beat-to-beat measures. Figure 10, published many years ago by Hewlett Packard, demonstrates that each beat can generate over 20 of these types of measurements. Most of these parameters are derived from the morphology of the waves and most do not relate to an actual physiological event. For example, the durations of the individual components of the QRS-complex, for example, Q_D, R_D, S_D, are not used by physicians when reading ECGs. However, such measurements may provide the software algorithms with values which aid in discriminating various abnormal QRS-complexes.

The second phase of the interpretation is generating diagnostic statements, samples of which are shown in the top portion of Fig. 9. For the most part these phrases are given the most attention by the physicians. They are usually rule-based outputs derived from the measurement matrix. In almost all cases these diagnostic statements are overread by the physician and changed if necessary.

The automated process is performed on almost all ECGs obtained in clinics and hospitals. Perhaps one of the greatest advantages of these automated systems is the digital ECG database. These systems are considered ECG management systems which allow for a tremendous saving in space needed for ECG storage, instantaneous recall of older recordings, and comparing serial changes in the ECG to chart the progress of some diseases.

The Monitored ECG

There are two primary applications where a patient's ECG is continuously monitored. Intensive care units within a hospital often monitor the ECG of critically ill patients to observe the patient's rate and rhythm. When the patient is suspected of having a life-threatening arrhythmia, it is best to monitor the patient in an environment where a rapid response and therapeutic intervention can be lifesaving. The other application of the monitored ECG is in the ambulatory, outpatient setting. The patient's ECG can be monitored by a belt-worn device. In the 1950s Norman J. Holter (13), an American physicist, demonstrated that the ECG could be monitored while the subject was physically active. However, the technology of the day resulted in a very heavy backpack device weighing 85 lb and was impractical for routine use. Recording devices and their associated electronics and batteries eventually became small enough to allow for a belt-worn tape recorder using originally reel-to-reel technology, but now relying exclusively on cassettes. These tape devices still have many inherent limitation, such as poor noise figures, low dynamic range, and limited frequency response. If one is merely recording the patient's rate and rhythm, then the tape technology is adequate. However, high-resolution ECGs might also be useful when obtained from the ambulatory patient, and the tape technology is definitely limited for this application.

Newer digital recorders are currently available whereby the ECG is digitized and stored in either high density memory chips or on actual hard disk drives. Current versions of the latter have removable drives with capacities exceeding 500 Mbytes. Depending on the application, the ECG may be sampled between 250 Hz and 1000 Hz. Originally, only one ECG lead was recorded on tape-based systems, but the new systems are not limited by the poor frequency response of tape systems or the physical size constraints of magnetic recording heads. With digital systems the number of simultaneous (or near simultaneous) recordings is not particularly limited, but three or four ECG leads are a practical number. Electrode positions for these monitored leads do not follow the conventions of the 12-lead ECG and are often similar to the bipolar limb leads, where the electrodes are placed a few inches apart over the chest, creating several lead fields through the heart.

For hospital monitoring, where the patient is being evaluated for a critical cardiac condition, only a few leads are recorded, but a full 12-lead ECG is periodically recorded. In some cases the patients, although not acutely ill, are given the freedom to walk about the hospital and their ECG is telemetered via radio frequencies by an antenna/receiver net-

work. In such cases the goal is to monitor the patient's rhythm in "normal" activities.

The massive amount of ECG information obtained during continuous monitoring is overwhelming. In the hospital approach, the ECG signals are usually fed to a large system of monitor screens where specially trained technicians view the actual recording of 10 to 50 patients. In conjunction with computer-based software the high risk situations are quickly identified with appropriate communication to the medical staff, for example, "code blue." This is not the case with outpatient monitors where the patient returns to the hospital 1 or 2 days later. Then the entire record is scanned with an interactive software analysis program. Often just a compressed printout of the continuous ECG can be quickly inspected for an abnormal rhythm. An example of this "full disclosure" mode is shown in Fig. 11. A 7.5 min recording is shown in this format. Note that the first several minutes have a normal rhythm (there are several other abnormalities in this tracing, but they are beyond the scope of this article). The abnormal beats begin to appear in groupings of two or three. A condition known as nonsustained ventricular tachycardia appears in the fourth trace from the bottom.

The event recorder is an extension to ambulatory recordings. In this case the patient wears a recorder for many days or even weeks. When the patients experience a symptom, such as chest palpitations or dizziness, they push an event button on the recorder which causes the recorder to save 1 min to 2 min of data before and after the event. The patient can call the physician office and transmit these data via a modem for rapid interpretation. The most advanced version of the event recorder is an implantable device that monitors the ECG for months or years and uses special monitoring software to record suspicious events without patient activation. This type of unit can be interrogated at regular intervals over the phone or during regular visits.

The Exercise ECG

The 12-lead ECG is obtained while the patient is resting. The monitored ECG is obtained either during an evolving disease

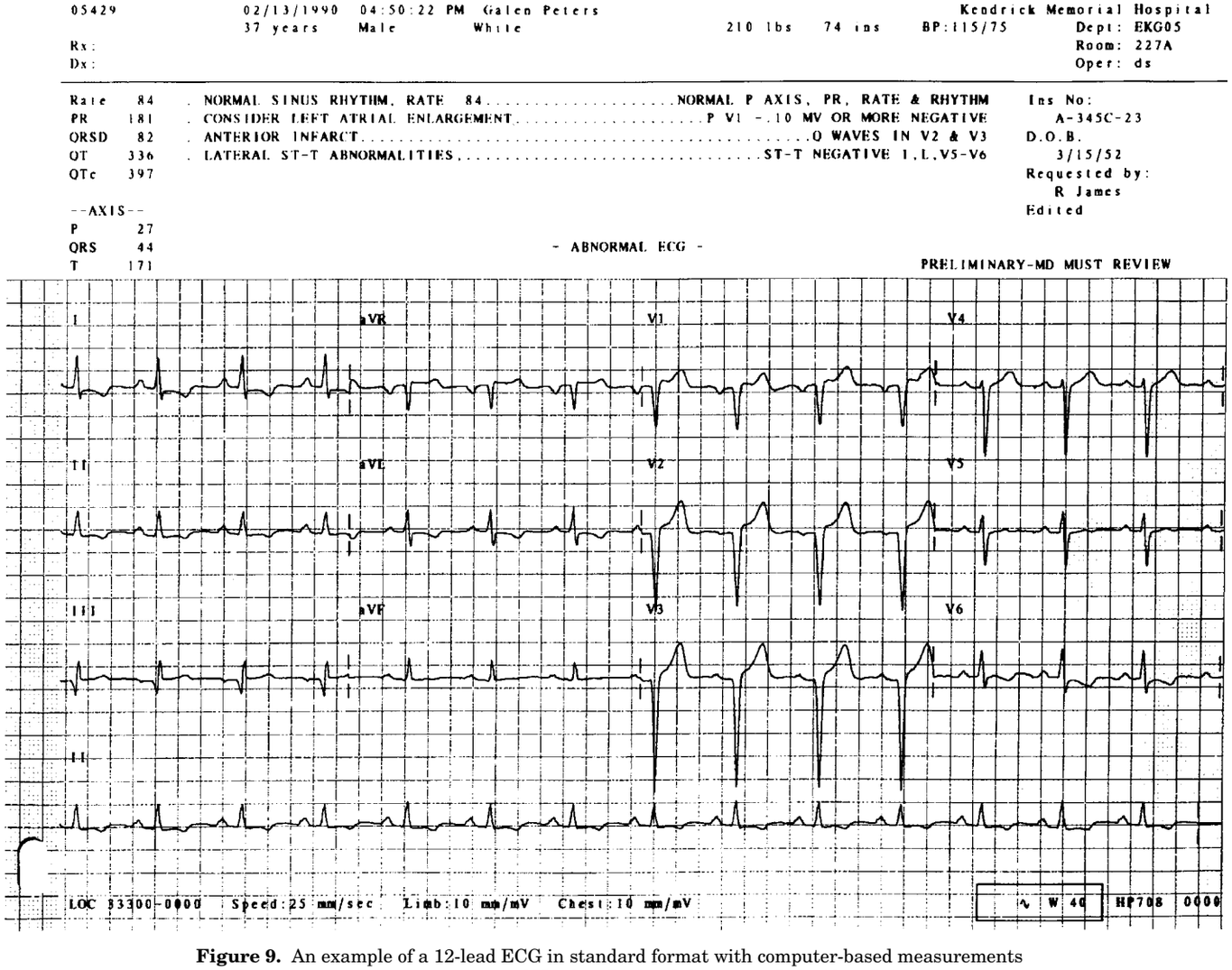

Figure 9. An example of a 12-lead ECG in standard format with computer-based measurements and diagnostic statements (courtesy of Hewlett-Packard Company, Palo Alto, CA).

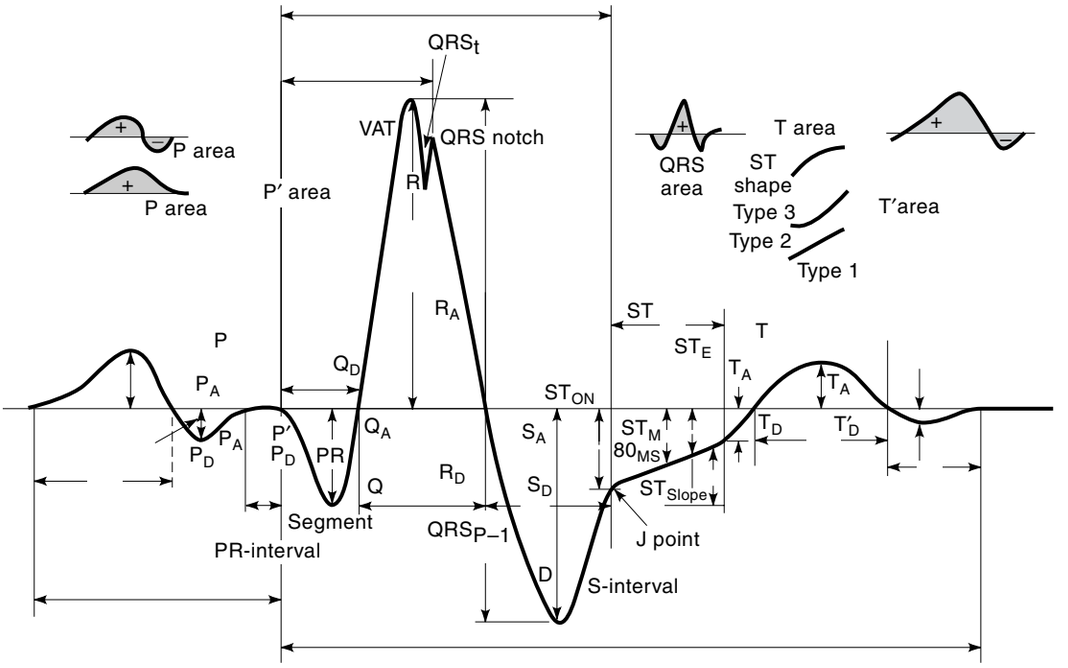

Figure 10. A stylized ECG showing the many amplitude, duration, and area measurements used to develop the measurement matrix for automated ECG interpretation (courtesy of Hewlett-Packard Company, Palo Alto, CA).

process or while the patient is undergoing routine activities. Evaluation of cardiac performance is done with a number of diagnostic tools, but an exercise stress test monitors the ECG prior to an exercise protocol (usually on a treadmill with adjustable speed and elevation) during the exercise period and finally during a warm-down period. Of particular interest is achieving a target heart rate which is usually based on the patient's age. The primary objective of the stress test is to monitor the ST-segment for small voltage changes on the order of at least 100 μV. This is a rather difficult measurement because of the amount of motion artifact resulting from the very active patient on the treadmill. Great attention must be paid to stable electrode placement on the chest. A number of computer algorithms can reduce the noise.

The High-Resolution ECG

Computer processing of the ECG has its origins in aiding or replacing the physician in making tedious measurements. Elucidating the standard waves (P-wave, QRS-complex, T-wave) was the primary aim. Once in the digital domain, a number of digital signal processing techniques are performed on the ECG, particularly in the realm of noise reduction and signal enhancement. This allows visualizing very low level signals ($\sim$1.0 μV) from such sources as the bundle of His and the left and right bundle branches. These structures are depolarized during the PR-segment (see Fig. 4) when no other cardiac signals are considered present. In fact, in traditional ECG theory the PR-segment is considered an isoelectric interval and is often used as a 0.0 V reference for the other waves. The initial computer-based approaches to record such a high-resolution ECG began in the early 1970s (14). Up to that time the His–Purkinje signals were recordable only by placing electrode catheters inside the heart and in very close proximity to the respective structures. The noninvasive recording of His–Purkinje signals was the advent of a new generation of ECG analysis. As the techniques for high-resolution ECG evolved perhaps the most clinically significant application was in recording so-called cardiac late potentials.

Cardiac late potentials typically arise from ventricular cells which surround a dead region of the heart caused by a heart attack. These bordering regions with surviving cells appear on the outer edge of the scar tissue and also permeate into the scarred region. It is possible that complete but circuitous pathways of viable cells can actually traverse the scar tissue. Such a matrix is often the site where life-threatening arrhythmias originate and are sustained. During normal heart rates partial activation of these arrhythmia pathways have been revealed, originally with electrodes in direct cardiac contact and then eventually by using high-resolution ECG techniques similar to those used to record His–Purkinje signals noninvasively (15). The activation of border zone cells is often delayed past activation of the normal ventricular cells because they are poorly conducting due to the heart attack. The artificially long pathways resulting from the mix of dead and surviving cells within and surrounding the infarct can also result in depolarizing signals which outlast the end of normal activation. These signals are not part of normal cardiac activation. The use of computer-based enhancement techniques has been the only way to identify and quantify them. It has been shown in hundreds of studies that the presence of these late potentials, after patients have had heart attacks, indicates that they have a very high risk of future life-threatening arrhythmias.

The primary method used to record both His–Purkinje and late potential signals is achieved by means of signal averag-

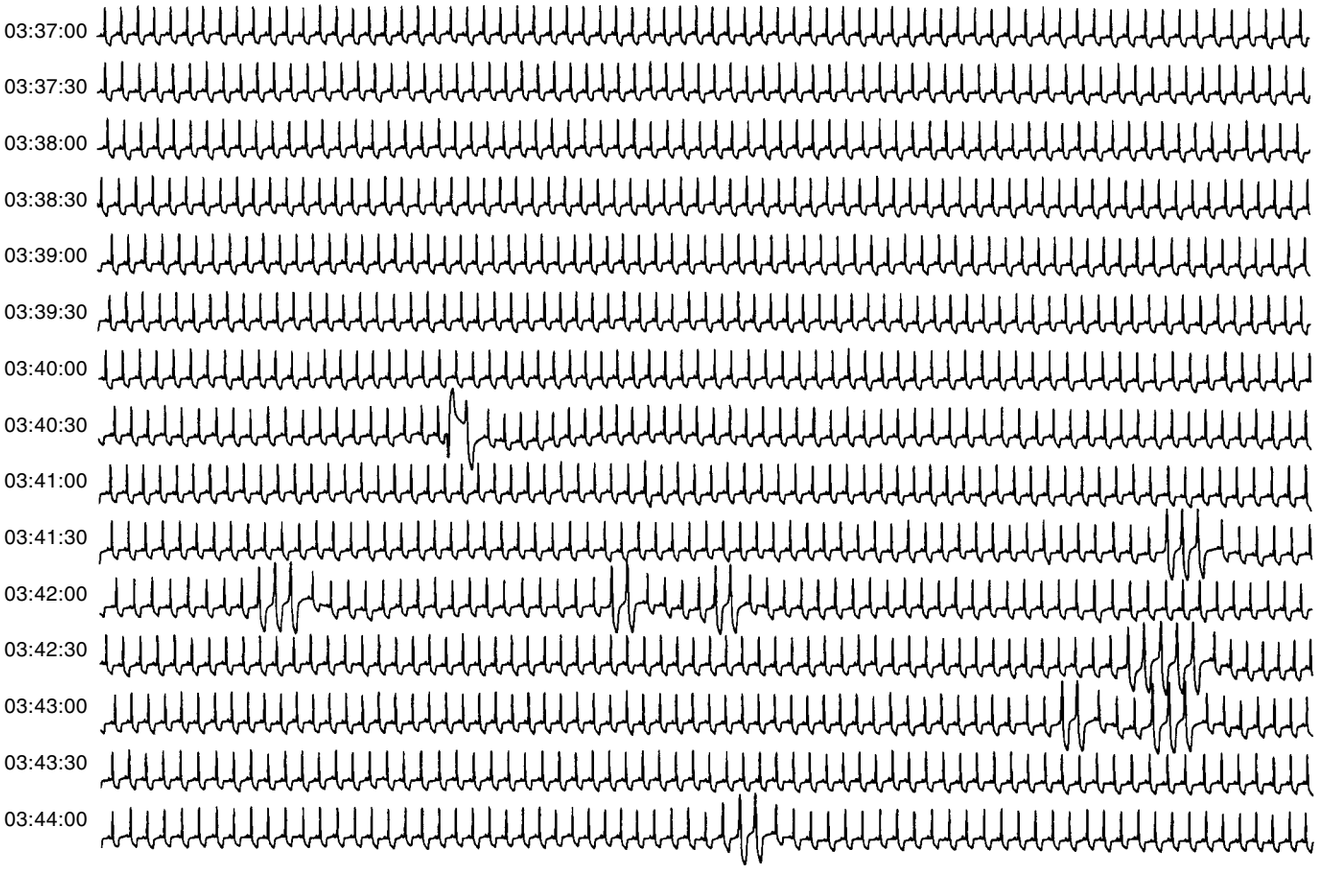

Figure 11. A portion of a full disclosure ECG from an ambulatory ECG. This mode of presentation, although significantly condensed, allows the trained reader to assess rhythm alterations rapidly. This is a very abnormal recording.

ing (16). One assumes that the signal of interest, which is very low level, repeats on a beat-to-beat basis. Also, the interfering noise (usually the signals associated with chest wall muscles depolarizing during breathing) is not temporally linked to the signal of interest and is random. Once digitized, the QRS-complex is detected very precisely so that the computer can finely align a window of data surrounding the QRS-complex and perform a point-by-point addition of each incoming beat. This averaging process is similar to one used by any scientist making a physical measurement. It is common to make several such measurements and to average them to get closer to the true value. The mathematics of signal averaging shows that under ideal conditions the signal-to-noise ratio increases by the square root of the number of measurements (beats). Thus if 100 beats are averaged, the noise decreases by a factor of 10.

Figure 12 best demonstrates the signal-averaged ECG using an example of cardiac late potentials. Figure 12(a) is an example of three ECG leads obtained from anatomically orthogonal leads called *X*, *Y*, and *Z*. These are bipolar leads, like leads I, II, and III, but the electrodes are placed on the chest along a set Cartesian axes with the heart at the origin. The voltage and timescale used in panel (a) are close to those used in the standard ECG. Panel (b) is a 300 ms window of the ECG approximately centered on the QRS-complex. The signal amplitude is five times greater than that used in panel (a). This single cardiac complex, however, is the result of averaging 200 cardiac cycles. One can actually observe small undulations at the end of and after the QRS-complex. These are late potentials, but they are still difficult to observe and quantify. One method used to elucidate overlapping signal components is to use a selective filter which reduces or increases the amplitude of certain spectral components of the signal (17). The bass and treble controls of a stereo amplifier are familiar to most as a means spectrally manipulating a signal. In the case of late potentials a special high-pass filter is used to reduce low frequency components and to pass the higher frequency components. The result of this is shown in panel (c). The QRS-complex appears as a very large multiphasic signal because of the very large amplification and the frequency-selective nature of the filter. The late potentials are clearly seen as the post QRS components. One commonly used presentation format is to combine the *XYZ* leads into a vector magnitude $(X^2 + Y^2 + Z^2)^{1/2}$. This is shown in panel (d), and the late potentials are shown in the shaded region. By making specific measurements on this filtered, vector magnitude, a number of parameters have been used to characterize the patient at high risk from life-threatening arrhythmias. The power of this method is significantly increased when used in conjunction with other clinical tests.

The Intracardiac ECG

There are many applications where electrical activity measured directly from the heart surface is the primary form of

data recorded for analysis. A modern electrophysiological evaluation of the heart relies on both the body surface ECG and direct recordings obtained from within the heart by electrode catheters. Such catheters are introduced into a leg or arm vein or artery and advanced, under fluoroscopic control, into the interior of one of the four chambers of the heart. An electrode catheter is an insulated set of wires bundled within a polyurethane sheath. The diameters of these catheters range from about 1.0 mm to 2.5 mm. As many as 16 wires may be in the total assembly with ring electrodes, exposed on the outer surface of the catheter, attached to each internal wire. In addition, there are usually structural internal wires used to stiffen the catheter. With a proper controller at the rear of the catheter, a trained operator can flex the catheter in a loop of almost 180°. Together with the torsional properties of the catheter, almost every point within the heart can be probed for electrical events. Direct contact recordings are called electrograms to distinguish them from body surface electrocardiograms.

Figure 13 is an example of a His bundle recording. The top two traces are leads II and V_6 of the ECG, and the bottom trace is the voltage difference from two electrodes on the indwelling electrode catheter. This internal view of cardiac activation combined with the His bundle electrogram has been called His bundle electrocardiography (18). Atrial activation on the catheter recording is called the "A" deflection and ventricular activation called the "V" deflection. The His bundle potential is the central "H" deflection. Because the catheter is located very close to the His bundle and AV node, it is assumed that the A deflection arises from atrial muscle tissue close to the AV node. When combined with the surface lead information, a number of new intervals can be obtained. These are the PA-, AH-, and HV-intervals. The PA-interval is a measure of atrial muscle activation time, the AH-interval is

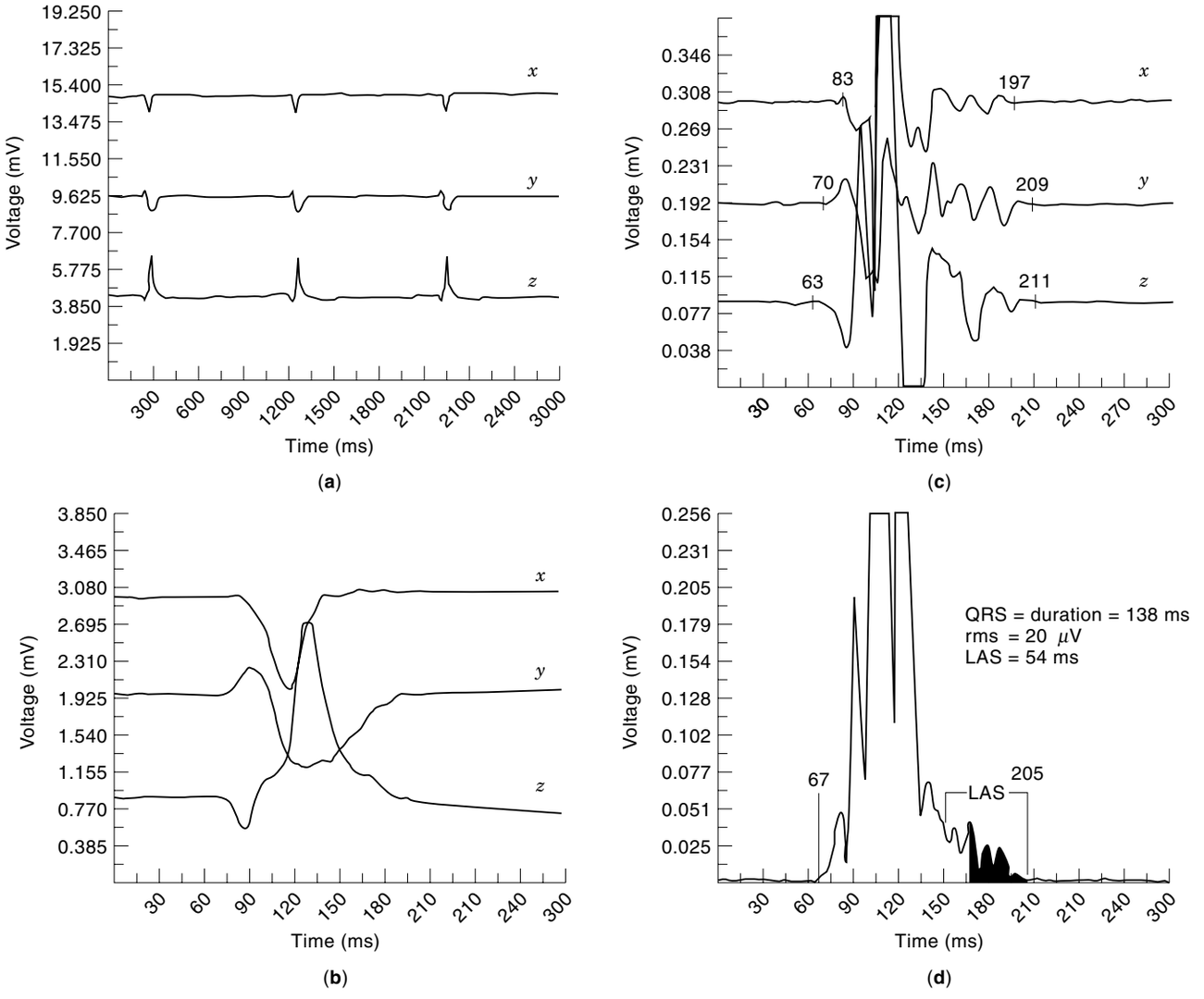

Figure 12. The high-resolution ECG derived by signal averaging the *XYZ* leads. Panel (a) is a 3s run of the normal scale *XYZ* leads. Panel (b) is a 0.3s window of the averaged *XYZ* leads with a fivefold increase in the voltage scale. Panel (c) is a 40 Hz high-pass version of panel (b). Panel (d) is the filtered vector magnitude derived from the *XYZ* with standard measurements indicated (see text). (Reprinted from the *Archives of Internal Medicine,* vol. 148, page 1862, 1988, Copyright 1988, American Medical Association.)

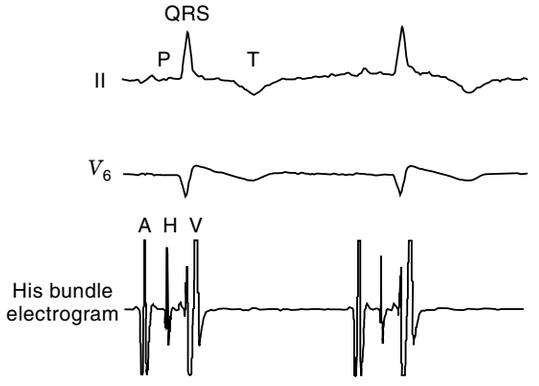

Figure 13. The top two traces are ECG leads II and V_6, and the bottom trace is a bipolar catheter recording, properly positioned inside the heart, showing the His bundle deflection (H), and intracardiac atrial (A) and ventricular (V) activity.

a measure of AV nodal activation time, and the HV-interval is a measure of the ventricular conduction system activation time.

The modern electrophysiological evaluation, or EP study, may involve as many as 64 individual recordings within the heart. In addition, current can be passed through these electrodes to stimulate the heart. A variety of atrial and ventricular stimulation protocols can be used which then allows the cardiac electrophysiologist to identify pathways and mechanisms involved in most forms of arrhythmias. Besides this diagnostic function, it is now possible to locate abnormal structures or regions of the heart which are critical to arrhythmogenesis. By passing high-energy, radio-frequency waves through one or more of the internal electrodes, it is possible to cauterize or ablate the suspect tissue without causing any widespread injury to the rest of the heart. In many forms of arrhythmias, this ablation therapy can produce a cure for the patient.

In addition to the EP study and ablation therapy, internal electrodes are the primary form of signal recording for both the cardiac pacemaker and implantable defibrillator. These devices sense cardiac activation from permanent indwelling catheters and deliver energy to the heart through them. In the case of the cardiac pacemaker these are low level shocks that maintain the patient's heart rhythm. In the case of the implantable defibrillator the device monitors the patient's rhythm until a serious or life-threatening arrhythmia occurs, and then a high-energy pulse is delivered to convert the rhythm back to normal. Both devices rely heavily on continuous monitoring of the cardiac signals obtained from internal catheter recordings using sophisticated implanted microprocessors and accurate means of signal detection and analysis.

CONCLUSION

The heart is a vital organ that generates an electrical signal which is a byproduct of the electrical triggering of the mechanical function of pumping blood. Recording and analyzing these electrical signals has a long history. The application of modern computer-based processing has automated this process and has also opened the way to many new and exciting methods for identifying patients at high risk from life-threatening arrhythmias and for enabling lifesaving devices.

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