Chapter 3

Cardiology

Electrocardiography

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The heart is a muscular organ responsible for circulating blood through the body’s circulatory system. The pumping action of the heart is coordinated by electrical activity that propagates through the heart’s four chambers. This same electrical activity makes it possible to noninvasively evaluate some aspects of heart function from recordings known as the electrocardiogram (ECG). The ECG is currently the fundamental technology used by physicians to evaluate the electrical activity of the heart and diagnose conditions such as abnormal rhythms (arrhythmias), abnormal conduction, damage due to heart attacks, and abnormal chamber sizes.

From a functional perspective, the basic operation of the heart consists of coordinated contractions of its four chambers: the right atrium, the left atrium, the right ventricle, and the left ventricle. A diagram is shown in Figure 3.1. The right atrium receives
deoxygenated blood, which then fills the right ventricle. The right ventricle then pumps this blood to the lungs via the pulmonary artery. The left atrium receives oxygenated blood from the pulmonary veins, which then fills the left ventricle. The left ventricle then pumps the oxygenated blood to all parts of the body via the aorta. The pumping of the heart occurs with rhythmic contractions that occur at rates typically between 60 and 80 beats/min. Heart rates vary depending on need (e.g., higher heart rates with physical activity but lower rates at rest).

On the cellular level, the heart is made up of electrically active cells called myocytes. Each cardiac myocyte in its resting state has a potential difference from the inside to the outside of the cell due to concentration gradients of sodium, calcium, and potassium ions. Activation of a cardiac myocyte is invoked either by an activation of a neighboring cell, through an applied electrical impulse, or even by an automatic mechanism. The response of an activated cell consists of inward and outward flows of the sodium, calcium, and potassium ions, which in sum produce a rapid depolarization of the intracellular potential and the subsequent gradual repolarization. The typical waveform of the intracellular potential during this process, known as the action potential, is shown in Figure 3.2. The refractory period in which the cardiac cell is not excitable after activation is related to the duration of the action potential. The refractory period tends to decrease with an increase in heart rate.

In a normal heart rhythm, a cluster of specialized pacemaker cells in the right atrium, known as the sinus node, initiates each heart beat. Activation wavefronts then propagate into the right and left atria, resulting in contraction of the atria. The atria and the ventricles are electrically connected by a specialized pathway called the atrioventricular node. Propagation through the atrioventricular node is slow compared to conduction in the atria and ventricles, allowing the atria time to contract before contraction of the ventricles is initiated.

The summed strength of the electrical activity from the large number of cardiac cells that make up the heart and the coordination of the electrical activity allows electric potentials to be detected using electrodes placed in contact with the wall of the heart or even

![Figure 3.2](image-url)
externally from the surface of the body. Recordings made from electrodes placed directly on the heart are known as electrograms, while body surface recordings of cardiac activity are known as ECGs.

The ECG during a normal sinus rhythm has three main components, as illustrated in Figure 3.3. The first sign of activity is the P-wave. This typically low amplitude hump reflects atrial depolarization. The next waveform, the QRS complex, occurs during the ventricular depolarization process. The third waveform is the T-wave, which reflects ventricular repolarization. The waveforms can vary in size, timing, and morphology, depending on the rate, rhythm, heart condition, and electrode locations.

3.2 ELECTRODES

The electrical activity of the heart has strong enough magnitude to be detected on the surface of the body. However, if one were to attempt to measure the electrical activity of the heart by simply touching the positive and negative ends of an oscilloscope by the left and right hands, no potentials relating to cardiac activity would be observed. The reason for this is that the oscilloscope requires an electrical current (i.e., a flow of electrons). In contrast, biopotentials from cardiac activity are the result of ionic currents. Thus, an interface between the oscilloscope probes and the skin that converts ionic currents to electrical currents is needed for the oscilloscope to display the body surface cardiac potentials. The same is true when an ECG recording system is used to record surface potentials.

Electrodes are transducers that convert the ionic currents to an electron current from which the ECG potentials can be amplified, recorded, and displayed. Electrodes are attached to the skin with an electrolyte gel serving as an interface between the electrode and the skin. The electron current is created by a chemical reaction between the electrolyte and the electrode.

There are two classifications of ideal electrodes: polarizable and nonpolarizable. With a polarizable electrode, the behavior of the electrode is similar to that of a capacitor. Current flowing through the electrode will charge the electrode causing “polarization.” The current responsible for charging (or discharging) the electrode is known as “capacitive current.” In a polarizable electrode, there is no “faradaic” current, which is the current
due to chemical reaction within the electrode, and no charge actually passes through the electrode–electrolyte interface. The potential of nonpolarizable electrodes, on the other hand, will not change from its equilibrium potential even with large current through the electrode. This behavior is attributed to the extremely fast electrode reaction (has an almost infinite exchange current density). Polarizable electrodes are thus more suitable for recording, while nonpolarizable electrodes are more suited for pacing and stimulation. Figure 3.4 shows the circuit equivalent of a biopotential electrode and the electrolyte interface. With a polarizable electrode, the resistance of \( R_d \) is infinite, and capacitor \( C_d \) is responsible for the polarization. With a nonpolarizable electrode, the capacitance of \( C_d \) is negligible.

3.3 LEAD SYSTEMS

Activation and repolarization of the heart is a 4D process. Therefore, multiple leads are needed to capture directional information of activation and repolarization. In this section, the common electrode configurations used for ECG recordings will be described.

3.3.1 12-Lead ECG

The 12-lead ECG is the most widely used configuration by cardiologists for diagnostic surface ECG evaluation. The 12-lead ECG tracings in this configuration are derived from electrodes placed in 10 standardized locations. The 12 leads are also commonly classified into 3 categories: the limb leads, the augmented limb leads, and the precordial leads. The limb leads (leads I, II, and III) are recorded from electrodes placed on the arms and legs, as shown in Figure 3.5a. The bipolar recording measuring the difference across the left arm (+) and the right arm (−) is designated as lead I. The bipolar recording across the left leg (+) and the right arm (−) is designated as lead II. Lead III is recorded between the left leg (+) and the left arm (−). The right leg is used as a reference for the amplifier circuitry. Lead III can be obtained by subtracting lead I from lead II. The vector representation of the limb leads known as Einthoven’s triangle is shown in Figure 3.5b.

The augmented limb leads (leads aVL, aVR, and aVF) are considered unipolar recording, because the positive electrode is recorded in reference to a combination of other electrodes. For aVL, the left arm potential is recorded in reference to the average of the right
arm and left leg potentials. For aVR, the right arm potential is recorded in reference to the average of the left arm and left leg potential. For aVF, the left leg potential is recorded in reference to the average of the right and left arm potentials. The augmented limb leads can also be derived from leads I and II with the following equations:

\[
\begin{align*}
    aVL &= \frac{-Lead I + Lead III}{2} = \frac{Lead I - Lead II}{2} \\
    aVR &= -\frac{Lead I + Lead II}{2} \\
    aVF &= \frac{Lead II + Lead III}{2} = \frac{Lead II - Lead I}{2}
\end{align*}
\]

The combined vector representation of the limb leads and augmented limb leads is shown in Figure 3.5c. Propagation of cardiac potentials traveling in the direction of the vector of a lead will result in a positive change in amplitude in that lead. Propagation in the opposite direction will result in a negative change in amplitude. Propagation perpendicular to the lead vector will result in zero potential.

The precordial leads of the 12-lead system consist of leads V1, V2, V3, V4, V5, and V6. The precordial leads are unipolar leads that use a common reference known as Wilson’s central terminal. Wilson’s central terminal is essentially the average potential of the right arm, left arm, and left leg electrodes. This average is obtained by the
interconnection of the three electrodes with resistors. The positive electrodes of the precordial leads are placed in the following locations (Figure 3.6):

- V1: right fourth intercostal space
- V2: left fourth intercostal space
- V3: halfway between V2 and V4
- V4: left fifth intercostal space, midclavicular line
- V5: horizontal to V4, anterior axillary line
- V6: horizontal to V5, midaxillary line

V1, V2, and V3 are used to detect electrical propagation traveling in the anterior/posterior directions (i.e., front/back). V4, V5, and V6 are used to detect lateral propagation (i.e., left/right).

3.3.2 Mason–Likar Lead Configuration

Although the 12-lead ECG is considered the standard for diagnostic ECG, the limb lead locations on the arms and leg make the recordings susceptible to motion artifact. Not only are the limb leads and augmented limb leads affected by movement of the arm and legs, the precordial leads are also affected, because Wilson’s central terminal is derived from the limb electrodes. In the Mason–Likar lead configuration, the arm electrodes are moved to the shoulders, and leg electrodes are moved to the hips to minimize movement of these electrodes. Thus, this system is often used in ambulatory (Holter) monitoring and exercise testing. Precordial leads remain in the standard positions if used with the Mason–Likar configuration.

3.3.3 Expanded 12 Leads

In the standard 12-lead system, the majority of the unipolar precordial leads are positioned in the left torso. Although not commonly used, the mirrored locations of V3, V4, V5, and
V6 on the right side of the torso are labeled lead V3R, V4R, V5R, and V6R. Additionally, V7 (the posterior axillary line), V8 (below the scapula), and V9 (paravertebral border) can be used to record from the posterior chest wall. These additional leads may be useful in the detection of myocardial ischemia and/or infarction.

3.3.4 Orthogonal Lead Configuration

Orthogonal lead systems were developed for the ability to record and display vectorcardiograms (VCGs). VCGs are created by plotting two or three orthogonal leads as a function of the other. The result is a display that shows the progression of dipole vectors as a series of loops. The magnitude of the vectors allows evaluation of the ECG waveform independent of the specific cardiac axis of the patient. Although the VCG is rarely used today in clinical practice, orthogonal ECG lead systems are still used in specialized ECG tests, such as signal-averaged ECG and microvolt T-wave alternans testing.

The Frank lead system, the most widely used orthogonal lead configuration, has the following seven electrode locations (shown in Figure 3.7a):

- A: left midaxillary line
- E: midsternum
- C: anterior left chest wall midway between A and E
- I: right midaxillary line

![FIGURE 3.7](image-url) (a) Frank lead electrode placement. (b) Resistor network for lead derivation.
• M: mid-spine
• H: junction of the neck and torso posteriorly
• F: left foot

From these electrodes, the leads \(X\), \(Y\), and \(Z\) are derived by the following weighted difference equations:

- \(X = 0.610 \, (A - I) + 0.170 \, (C - I)\)
- \(Y = 0.345 \, (M - E) - 0.655 \, (H - F) + 0.345 \, (E - H)\)
- \(Z = 0.132 \, (A - I) + 0.372 \, (M - E) + 0.365 \, (M - C) + 0.132 \, (C - I)\)

The weighted differences can be achieved by using a resistor network as shown in Figure 3.7b or can be performed in software.

3.3.5 Lead Configurations for Bedside and Ambulatory Monitoring

Lead configurations for bedside and ambulatory monitoring have traditionally used reduced number of electrodes compared to the 12-lead ECG to save transmission bandwidth and recording space. Three-electrode and five-electrode systems are commonly used. The three-electrode system consists of a positive electrode, negative electrode, and a ground electrode to obtain one ECG channel. The electrodes can be configured to obtain the Mason–Likar equivalent of lead I, II, or III. Often a modified chest lead (MCL) is used (MCL1), which approximates a lead V1 signal by placing the positive electrode in the standard V1 location and the negative electrode on the left infraclavicular fossa. MCL1 will differ from the standard V1 due to the change in reference from the Wilson’s central terminal to the left infraclavicular fossa. With a five-electrode system, the complete frontal plane leads can be simultaneously obtained (I, II, III, aVR, aVL, aVF) in addition to one chest lead (V1).

Philips Medical Systems (Eindhoven, Netherlands) also has a five-electrode system known as the EASI configuration that uses electrodes A, E, and I from the Frank lead system, an electrode on the sternal manubrium (S), and a ground electrode that can be placed anywhere on the torso (G). The advantage of this system is the ability to derive Mason–Likar 12-lead ECGs with only five electrodes instead of 10 electrodes facilitating setup and storage particularly for telemetry and bedside monitoring. Figure 3.8 shows the location of the EASI electrodes and the transformation coefficients for the derivation of the 12-lead signal.

3.3.6 Body Surface Mapping

Obtaining ECGs with high spatial coverage obtained by large numbers of electrodes is known as body surface mapping. The high density of recordings has been shown to allow the detection of the presence and location of myocardial infarction in cases where it was not possible with the 12-lead ECG. Body surface mapping has not gained wide
acceptance. As such, there are no accepted standards regarding the number or locations of the electrodes. The PRIME ECG system by Heartscape Technologies (Bothell, WA) is one commercially available body surface mapping system that records from 80 electrodes placed in the front and back of the chest. The locations of the PRIME ECG electrodes are shown in Figure 3.9. Body surface mapping is also a topic of current research in which the surface potentials are backprojected onto either 3D MRI or CT images of the patient’s heart to predict the propagation pattern of cardiac activation. This technique is known as electrocardiographic imaging.

3.4 AMPLIFIERS

The potentials recorded on the body surface have amplitudes on the order of 1 mV. A combination of amplification and filtering is needed to obtain diagnostic quality tracings. A block diagram of the typical components of an ECG amplifier circuit is shown in Figure 3.10.

![Transformation coefficients](image)

<table>
<thead>
<tr>
<th>Lead</th>
<th>ES</th>
<th>AS</th>
<th>AI</th>
</tr>
</thead>
<tbody>
<tr>
<td>I</td>
<td>0.026</td>
<td>-0.174</td>
<td>0.701</td>
</tr>
<tr>
<td>II</td>
<td>-0.002</td>
<td>1.098</td>
<td>-0.763</td>
</tr>
<tr>
<td>III</td>
<td>-0.028</td>
<td>1.272</td>
<td>-1.464</td>
</tr>
<tr>
<td>aVR</td>
<td>-0.012</td>
<td>-0.462</td>
<td>0.031</td>
</tr>
<tr>
<td>aVL</td>
<td>0.027</td>
<td>-0.723</td>
<td>1.082</td>
</tr>
<tr>
<td>aVF</td>
<td>-0.015</td>
<td>1.185</td>
<td>-1.114</td>
</tr>
<tr>
<td>V1</td>
<td>0.641</td>
<td>-0.391</td>
<td>0.080</td>
</tr>
<tr>
<td>V2</td>
<td>1.229</td>
<td>-1.050</td>
<td>1.021</td>
</tr>
<tr>
<td>V3</td>
<td>0.947</td>
<td>-0.539</td>
<td>0.987</td>
</tr>
<tr>
<td>V4</td>
<td>0.525</td>
<td>0.004</td>
<td>0.841</td>
</tr>
<tr>
<td>V5</td>
<td>0.179</td>
<td>0.278</td>
<td>0.630</td>
</tr>
<tr>
<td>V6</td>
<td>-0.043</td>
<td>0.431</td>
<td>0.213</td>
</tr>
</tbody>
</table>

FIGURE 3.8 EASI lead placement and transformation coefficients for approximation of the 12-lead ECG.

Front view Back view

FIGURE 3.9 Electrode locations for the PRIME ECG body surface mapping system. Seven electrodes on the left side and six electrodes on the right side not shown.
The first component receiving the signals from the electrodes is the amplifier protection circuit. Diodes are often used to limit the effect of high voltage/current, such as in the cases of electrostatic discharge or defibrillation shock. Some ECG systems have lead failure detection circuits that operate at this stage as well. The next block is the lead selector circuit. This circuit is usually controlled by the microprocessor to select the combination of electrodes required to be amplified. Next is the preamplifier where initial amplification is performed. The preamplifier should have high common-mode rejection and input impedance. The next stage is the isolation circuit. For ECG machines running on AC power, the isolation circuit protects both the patient and the ECG machine from possible high voltages. Isolation circuitry is not needed for battery-powered recorders such as Holter monitors. The driven right leg circuit sets the right leg as the reference signal. Finally, the driver amplifier performs the final amplification as well as band-pass filtering before analog-to-digital conversion is performed. International guidelines recommend band-pass filtering with cutoff frequencies of 0.05–150 Hz and zero-phase distortion. Amplification is performed to achieve an amplitude resolution on the order of 10 µV or better prior to analog-to-digital conversion.

3.5 DISPLAYS

ECG recordings have traditionally been displayed with either oscilloscope-like tracings or printed directly to paper. Figure 3.11 shows an example of 12-lead ECG tracing in a typical clinically used format. The first three rows contain 2.5 s snapshots of each of the 12 leads. The fourth row consists of a 10 s rhythm strip of lead II. The ECGs are plotted on a grid pattern consisting of larger 5 by 5 mm boxes, which are in turn made up of smaller 1 by 1 mm boxes. The rectangular waveforms preceding each of the four rows are calibration pulses, which indicate the time scale (width of 200 ms) and amplitude scale (height of 1 mV). In this example, one big box (5 mm) equals 200 ms, and two big boxes (10 mm) equal 1 mV. It is important to examine these calibration pulses as ECG machines do allow alterations in both the time and amplitude scaling of the ECG.

Real-time displays present ECG waveform data as they are being collected. Real-time displays often display ECGs alongside heart rate measurements or other real-time physiologic measurement. The real-time waveforms can be either continuously scrolling or updating every few seconds at a determined sweep speed (typically 2–5 s).
3.6 ARRHYTHMIAS

Arrhythmias are abnormal heart rhythms with characteristics that differ from rhythms, which originate from the sinus node and conduct normally through the atria, atroventricular node, and ventricles. Figure 3.12a shows an illustration of an ECG during normal sinus rhythm. The ECG is an important tool for the evaluation of many types of arrhythmias. Table 3.1 shows arrhythmias that can be detected via automatic algorithms processing the digital ECG waveform. Illustrations of ECG manifestations of different arrhythmias are shown in Figures 3.13 and 3.14.

3.7 AUTOMATIC DIAGNOSIS

Automatic algorithms utilizing digital signal processing are capable of providing a plethora of ECG measurements and diagnoses. These measurements can include interval and amplitude measurements that can be confirmed by a physician with calipers. Automated algorithms can also involve rhythm and conduction classifications, which are then confirmed by a trained physician. More sophisticated signal-processing algorithms can also be used to extract information from the ECG that cannot be easily determined or quantified by humans. The following are examples of ECG measurements that can be obtained automatically.

3.7.1 QRS Detection

Robust QRS detection is required not only for the measurement of heart rate but is the basis for almost all automatic ECG detection algorithms. QRS complexes are characterized
by high amplitude and steep slopes. Both amplitude and slope can be used for detection, but slope usually offers better discrimination from the T-wave. Figure 3.14 compares amplitude versus slope-based QRS detection in an ECG where high-amplitude T-waves are present. QRS detection methods almost always include an assumption of a minimum refractory period to be used as a blanking period during which subsequent beats cannot be detected. Cross-correlation using a QRS template can be used to reduce timing jitter compared to relying on the maximum amplitude or slope alone as the fiducial point. Template comparisons can also be used in the detection and classification of premature ventricular complexes.

The time period between QRS complexes is known as the RR interval. Ventricular (heart) rate in beats per minute can be calculated by dividing 60,000 by the mean RR interval (in milliseconds) over a time period.

3.7.2 QRS Duration

The measurement of the duration of the QRS complex is used to evaluate intraventricular conduction. Short QRS durations usually reflect normal and synchronized conduction through the
TABLE 3.1 Examples of Common Arrhythmias and Abnormalities That Can Be Determined from ECG Tracings

<table>
<thead>
<tr>
<th>Arrhythmia/Abnormality</th>
<th>Description</th>
<th>ECG Characteristics</th>
</tr>
</thead>
<tbody>
<tr>
<td>Sinus bradycardia</td>
<td>Sinus rhythm with very slow heart rates (&lt;60 beats/min).</td>
<td>Mean interval between QRS complexes is greater than 1 s (Figure 3.12b).</td>
</tr>
<tr>
<td>Sinus tachycardia</td>
<td>Sinus rhythm with very fast heart rates (&gt;100 bpm).</td>
<td>Mean interval between QRS complexes is less than 600 ms (Figure 3.12c).</td>
</tr>
<tr>
<td>First degree AV block</td>
<td>Long conduction times through the atrioventricular node.</td>
<td>Long intervals between the P-wave onset and the QRS onset (&gt;200 ms) (Figure 3.12d).</td>
</tr>
<tr>
<td>Type 1 second degree AV block (Wenckebach)</td>
<td>During a stable sinus rhythm, there is progressively longer conduction times through the AV node until conduction is blocked for one sinus beat, after which conduction times are reset and the pattern repeats.</td>
<td>The interval between the P-wave onset and the QRS onset progressively lengthens until there is a P-wave without a trailing QRS complex and T-wave, after which the pattern repeats (Figure 3.12e).</td>
</tr>
<tr>
<td>Type 2 second degree AV block</td>
<td>Sudden conduction block of a sinus beat without preceding lengthening of the AV nodal conduction time.</td>
<td>A P-wave is not followed by a QRS complex and T-wave, without preceding lengthening of the PR interval (Figure 3.12f).</td>
</tr>
<tr>
<td>Third degree AV block</td>
<td>No conduction from the atrium to the ventricle.</td>
<td>The P-waves and QRS complexes are completely dissociated, with an atrial rate that is faster than the ventricular rate; QRS complexes can result from escape beats initiated from AV junctional or ventricular sources (Figure 3.12g).</td>
</tr>
<tr>
<td>Premature atrial complexes</td>
<td>An early beat initiated from another atrial site other than the sinus node.</td>
<td>P-wave occurs earlier than sinus-node-initiated P-waves, often with a different morphology (Figure 3.13a).</td>
</tr>
<tr>
<td>Premature ventricular complexes</td>
<td>Early beat initiated from a ventricular site.</td>
<td>QRS complexes, often with different morphology, occur early without a preceding P-wave (Figure 3.13b).</td>
</tr>
<tr>
<td>Atrial flutter</td>
<td>Rapid atrial activity with continuous but repeated patterns due to a reentrant circuit around an anatomic obstacle.</td>
<td>P-waves are replaced with F waves, often with a continuous sawtooth or sinusoidal-like waveform, with consistent timing and morphology (Figure 3.13c).</td>
</tr>
<tr>
<td>Atrial fibrillation</td>
<td>Rapid and seemingly uncoordinated activation of the atria.</td>
<td>Discrete P-waves are replaced by an undulating baseline with changing timings and morphologies (Figure 3.13d).</td>
</tr>
<tr>
<td>Ventricular tachycardia</td>
<td>An abnormally fast rhythm (&gt;100 bpm) originating from the ventricles.</td>
<td>Monomorphic ventricular tachycardia has a stable rate, rhythm, and QRS morphology; polymorphic ventricular tachycardia has changing rate, rhythm, and QRS morphology (Figure 3.13e).</td>
</tr>
<tr>
<td>Ventricular fibrillation</td>
<td>A type of polymorphic ventricular tachycardia with seemingly disorganized activity in the ventricles.</td>
<td>Undulating baseline with changing amplitudes and timing; no discrete QRS complexes or T-waves are present (Figure 3.13f).</td>
</tr>
</tbody>
</table>
ventricles, whereas long QRS durations may indicate conduction block somewhere along the normal conduction sequence of the ventricle. Normal QRS duration ranges from 60 to 110 ms.

The measurement of QRS duration requires the detection of the beginning (onset) and the end (offset) of the QRS complex (Figure 3.15). Because the QRS complex is typically characterized by sharp deflections, automatic algorithms used to detect QRS onsets and offsets rely on the slope of the signal. Starting with the fiducial point obtained by QRS detection, the QRS onset can be detected by the closest point preceding the fiducial point where the QRS slope falls below a threshold for a given amount of time. The QRS offset is
similarly detected as the closest point following the fiducial point where the QRS slope falls below a threshold for a certain amount of time. The QRS offset is also known as the J-point.

3.7.3 QT Interval

The QT interval is a measure of the time period from the QRS onset to the end of the T-wave. QT intervals provide an estimate of the time required for the ventricles to repolarize following activation. Because the time to repolarize is dependent on the heart rate, a heart-rate-corrected QT interval (QTc) is often used. QTc is calculated by the following formula (Bazett’s formula):

\[
\text{QTc} = \frac{QT}{\sqrt{RR}}
\]

where

\( QT \) is in milliseconds
\( RR \) is in seconds

A QTc interval of greater than 450 ms for males and 470 ms for females is an indicator for vulnerability to ventricular arrhythmias. Other rate correction formulae have also been used.

Automatic measurement of the QT interval requires detection of the QRS onset as previously discussed and the T-wave offset. The end of the T-wave is a challenge to accurately detect as the tail slowly returns to baseline, as shown in Figure 3.15. Small differences in amplitude or slope thresholds can drastically affect where the end of the T-wave is marked. A common alternative algorithm marks the end of the T-wave by projecting the tangent of the maximum descending slope of the T-wave to the baseline. Where the projection
intersects with the baseline is considered the T-wave offset. The point of the maximum descending slope can be detected reliably, and using it provides a T-wave offset that is more stable than that obtained by fixed amplitude or slope thresholds.

3.7.4 PR Interval
The PR interval is the time period between the P-wave onset and the QRS onset (Figure 3.15). The PR interval is used in the evaluation of atrioventricular node conduction and to evaluate heart block. The P-wave onset is automatically detected by the presence of an isoelectric baseline for a certain amount of time preceding the P-wave. During fast heart rates, the P-wave may be masked by the end of the T-wave of the preceding beat.

3.7.5 ST Height
The portion of the ECG between the QRS offset and the beginning of the T-wave is known as the ST segment (Figure 3.15). The ST segment reflects the early portion of repolarization of the ventricle. ST-segment elevation can be used as an indicator of occurring or prior myocardial infarction (i.e., heart attack) where portions of the ventricle are damaged or distressed due to the lack of blood supply from a blocked blood artery. ST elevation of more than 1–2 mm (0.1–0.2 mV) in two or more contiguous leads is an indicator of acute myocardial infarction. ST height can be measured automatically by determining the amplitude of the ECG relative to the isoelectric line at points at or after the QRS offset.

3.7.6 QRS Axis
The polarity of the QRS complex in the leads that define the frontal plane can indicate whether the ventricular activation sequence is propagating in the expected direction. The direction of the QRS complex is known as the QRS axis. The normal QRS axis ranges from −30° to +90°. With a normal axis, lead I has a positive deflection, and lead aVF or II generally has positive deflections. QRS axis of −30° to −90° indicates left axis deviation (lead I is positive and aVF is negative), while right axis deviation is defined from +90° to +150° (lead I is negative and leads II and aVF are both positive). Left or right axis deviation could indicate an abnormality in ventricular conduction.

3.8 ADVANCED ECG SIGNAL PROCESSING
Most of the interval, amplitude, and axis measurements described earlier can be done not only with automatic algorithms but can also be performed manually with calipers. Some ECG signal-processing methods have been developed that offer diagnostic information that cannot be easily obtained through manual analysis.

3.8.1 Median Beat Analysis
Noise in the ECG due to electromyogram artifact, motion, or poor electrode contact can make any of the previously described measurement difficult, either manually or automatically. Noise is a particular issue when ECGs are recorded in an ambulatory setting or for the purposes of an ECG stress test. Filtering can be used to improve the signal-to-noise ratio of the ECG, but overfiltering can cause distortion of the waveform. Median
templates of ECG waveforms are a composite of several beats. With a normal rhythm and a stable heart rate, most waveforms of each beat should be very similar to each other. Noise, on the other hand, is generally uncorrelated with ECG activity. Therefore, a composite of several beats will give us a truer sense of what the waveform should look like without noise.

The process of creating a median beat is illustrated in Figure 3.16. First, QRS detection is performed as previously described. Next, fixed length windows, enough to encompass each beat from the beginning of the P-wave to the end of the T-wave, are obtained. All the windows are then aligned with each other with the maximum slope of the QRS used as the fiducial point. A median value for each sample corresponding to a point in the cardiac cycle is obtained and used as the representative amplitude for that time point. The median beat is obtained when all median values for each point in the cycle are calculated.

3.8.2 Pacemaker Detection
With interpretation of the ECG, it is important to know if a pacemaker is controlling the heart beats. Because the output pacing stimuli of pacemakers have very narrow pulse width (generally on the order of 0.5 ms), ECG recording devices with standard sampling rates of 1000 Hz and less are not adequate to reliably capture the pacing activity (sampling at 1000 Hz is every 1 ms; a typical ECG sampling rate is 250 Hz or every 4 ms). ECG systems with pacemaker detection features have specialized hardware and software that detect the pacemaker impulses during acquisition. The process usually includes band-pass filtering at a high-frequency band, an analog comparator circuit, and high sample rate analog-to-digital conversion. The times of the detected pulses are then saved as annotation on the lower sample rate ECG.

3.8.3 Heart Rate Variability
Heart rate variability reflects the modulation of beat-to-beat firing rate of the sinus node by the autonomic nervous system. The autonomic nervous system consists of two arms: the sympathetic nervous system that controls the body’s “fight and flight” response and the parasympathetic nervous system that controls the body’s “rest and recovery” response.
Increase in sympathetic neural activity results in an increase in heart rate, while increase in parasympathetic neural activity decreases heart rate. The magnitude of resting heart rate variability is largely reflective of parasympathetic neural activity and/or its respiratory modulation. However, the relative effects of sympathetic and parasympathetic neural activity will determine the spectral profiles of the heart rate variability signal. Low heart rate variability has been shown to have prognostic value in determining risk for sudden cardiac death.

All measures to quantify heart rate variability require QRS detection, calculation of the intervals between the R-waves (RR intervals), and the rejection of RR intervals due to nonsinus node activity. The resulting data are known as NN intervals. There are several common heart rate variability measures that are used. In the time domain, these include the standard deviation of NN intervals (SDNN), the root mean square of successive differences (RMSSD), and the percentage of NN intervals with greater than 50 ms change from the preceding NN interval (pNN50). Frequency domain measures that have also been proposed as different frequency bands are thought to reflect different autonomic processes. Frequency domain parameters are obtained by first interpolation of the NN intervals to obtain a signal with a constant sampling rate (usually 2 Hz). Linear detrending of the resampled signal is then performed. The power spectrum is then obtained. The power in a high-frequency band (0.15–0.4 Hz) has been shown to reflect parasympathetic modulation of heart rate. A low-frequency band (0.04–0.15 Hz) has been shown to reflect both sympathetic and parasympathetic modulations of the heart rate. An example of NN intervals in the frequency domain is shown in Figure 3.17.

![Figure 3.17](image_url)

**FIGURE 3.17** Example of a 5 min NN interval series at rest and the corresponding power spectrum. The peak at 0.24 Hz corresponds to the respiratory modulation of heart rate.
3.8.4 Signal-Averaged ECG

Signal-averaged ECG is a technique used to identify patients with “late potentials” that occur due to slow conduction through infarcted areas of the ventricle. These late potentials occur at the end of the QRS complex and may indicate vulnerability to ventricular arrhythmias. Because of the low amplitude of the late potentials, specific signal-processing operations consisting of a combination of signal averaging and filtering are used. Signal-averaged ECG requires several minutes of resting ECG recording using an orthogonal lead system with a sampling rate of 1 kHz. Signal-averaged (mean) waveforms of the X, Y, and Z leads are obtained in a similar process used in the median beat analysis described previously. After signal averaging, the noise levels are effectively reduced by an order of magnitude. High-pass filtering with cutoff frequency of 30 Hz is then performed. The vector magnitudes of the filtered X, Y, and Z signal-averaged beats are then calculated. This process is illustrated in Figure 3.18. Using the vector magnitude, the following measures are typically used to detect the presence of late potentials: filtered QRS duration (>114 ms), root-mean-square voltage in last 40 ms of the QRS (<20 µV), and signal duration in terminal QRS that is <40 µV (>38 ms). Patients meeting two or three of these criteria have been shown to be at increased risk for sudden cardiac death.

3.8.5 T-Wave Alternans

At high heart rates, the T-wave can exhibit alternating beat-to-beat patterns of A to B to A to B, where A and B each represents a specific amplitude, duration, or morphology of the T-wave. Detection of these “T-wave alternans” patterns at relatively low heart rates has been shown to be a risk predictor of cardiac arrhythmias. Because the beat-to-beat changes...
often are in the order of microvolts, detection of alternans is not easily discerned visually and requires specialized signal processing.

Two prominent methodologies have emerged for the analysis of T-wave alternans that are used in the clinical setting. The first is a method developed by Cohen and colleagues that measures beat-to-beat alteration using the frequency domain. If a beat-to-beat sequence of T-wave amplitudes is transformed into the frequency domain, a prominent peak at the 0.5 cycles per beat frequency will be present in the power spectrum if alternans is present. If there is no T-wave variation or if the variation is not in an ABAB alternating pattern, the power at the 0.5 cycles per beat frequency will not be significantly different from the nearby frequencies. Figure 3.19 shows an example of a power spectrum with alternans present. A parameter called the “k-score” is used to determine if the alternans magnitude is significantly greater than the nearby “noise” frequencies. The k-score is calculated as the difference between alternans magnitude measured at the 0.5 cycles per beat frequency and the mean noise magnitude measured in the 0.43–0.47 cycles per beat band, which is then divided by the standard deviation of the noise magnitude. A k-score greater than 3 is considered significant alternans. Patients having significant alternans during a stress test at heart rates less than 110 beats/min have been shown to be a risk for ventricular arrhythmias. A commercial system performing this test is currently sold by Cambridge Heart (Bedford, MA).

**FIGURE 3.19** Simulation of alternans with added noise. (a) Raw ECG signal. (b) Measured T-wave amplitude. (c) Power spectrum of the amplitude values with a large peak at 0.5 cycles per beat indicating alternans. (d) Superimposed plots of the averaged even and odd beats. Separation occurs at the T-wave.
A second method to measure T-wave alternans is the “modified moving average” developed by Nearing and Verrier. This method is a time-domain approach that performs separate moving averages for even and odd beats. To account for noise and outliers, the algorithm limits the effect of any one beat on the calculations by using nonlinear update factors. If a certain amount of separation between the even and odd beats is detected, alternans is considered to be present. Patients with separation of more than 65 μV have been shown to be at high risk for sudden cardiac death. Figure 3.19 shows an example of an ECG tracing and the superimposed even and odd averaged beats. The test is available as part of the GE Healthcare’s (Waukesha, WI) Holter monitoring analysis software.

3.9 AMBULATORY MONITORING

The evaluation of the ECG during resting or stationary conditions is limited because of its short duration and typical absence of symptoms at the time of the recording. Ambulatory monitors can be used to monitor or record ECG activities for long periods of time, while the patient follows a normal daily routine. The likelihood of capturing transient ECG events is thus improved when these monitors are used.

3.9.1 Holter Monitoring

Holter monitoring involves using a portable ECG recording device that can continuously record ECG over 24 h or longer. Holter monitors commonly record ECGs from three to five electrodes with leads connecting to the belt-worn monitor. Twelve-lead recording systems are also available. The leads are often taped to the body for extra security. Holter monitors also feature an “event” button that allows the patient to document the time of any symptoms, which can then be correlated with ECG to determine whether the symptom is associated with a cardiac arrhythmia.

Early versions of Holter monitors were battery-powered cassette tape recorders that stored amplified ECG waveforms on the tape in analog form. As solid-state (“flash”) memory has become cheaper and more widespread; almost all Holter monitors currently store digital data. It is also common that data compression on the waveforms is performed prior to storage to save storage space. However, because the cost of flash memory is continuously decreasing, compression is becoming less necessary. Digital Holter recorders also have the advantage of not having any motorized parts, thus allowing greater battery life.

Although Holter monitors primarily function as a recording device, some monitors may have limited real-time capabilities such as lead checking and heart rate measurement. However, most processing occurs after the conclusion of the recording period when the signals are transferred to a workstation containing the analysis software. Some clinical information that is obtained in the analysis of Holter recordings includes QRS morphologies, heart rate, heart rate variability, intervals (PR, QRS duration, QT, etc.), and arrhythmia detection. Manual overreading is necessary with Holter monitoring because of the susceptibility of ambulatory recordings to noise. Motion and tremor artifacts are common sources of interference of the ECG signal. Thus, time segments with high levels of artifact must be excluded to prevent false reading of heart rate and other measures.
Some cardiac events occur too infrequently to be detected with even 24 or 48 h Holter monitoring. Recording longer continuous ECG for such situations may not be practical, not only from the standpoint of data storage but also for the amount of data that would need to be analyzed. Thus, an alternative to the Holter monitor is the event recorder.

### 3.9.2 Long-Term Ambulatory/Event Recorders

Event recorders are another form of ambulatory monitors. Event recorders differ from Holter monitors in that the ECG recordings are stored only if the patient starts the monitor when symptoms occur or if the device automatically detects an arrhythmia. Because the device is not recording continuously, event monitors are usually smaller in size than Holter monitors and can record for longer periods of time (30 days) because of reduced power requirements. The physician can then use the collected data to determine whether an arrhythmia was documented. Long-term continuous recorders are also available.

### 3.9.3 Implantable Loop Recorders

Holter and event monitors record and store ECG signals from surface electrodes. Implantable loop recorders are small devices that are implanted under the skin, usually beneath the collar bone. These devices record single-lead ECGs from electrodes that are on the case of the recorder. The loop recorder buffers a short amount of ECG at a time. The patient places a handheld activator over the area where the loop recorder is implanted when the patient senses symptoms. The recorder then transfers the ECG from the buffer to another memory location that can be accessed later. The loop recorders can continue to operate for up to 3 years after which they can be removed.

One application for these devices is to monitor patients for episodes of atrial fibrillation. These episodes can be both symptomatic and asymptomatic. Thus, a robust ECG signal processing is required to accurately detect these episodes. These recorders typically detect atrial fibrillation via the response of the ventricle, which is usually characterized by QRS complexes that are highly irregular in timing but similar in morphology. Medtronic (Minneapolis, MN) and St. Jude Medical (St. Paul, MN) both offer implantable loop recording devices. Loop recording is also performed in modern implantable pacemakers and defibrillators.

### 3.10 TELEMETRY

Telemetry is a system in which ECG recordings (and other physiologic measurements) are wirelessly transmitted to a separate monitor for real-time display and monitoring. The setup on the patient is similar to that of a Holter monitor. The ECG electrodes are connected to a portable wireless transmitter by leads. The transmitter contains amplifiers and modulators for radio-frequency transmission. In current systems, the transmitter performs analog-to-digital conversion before transmission using a digital wireless protocol.

Inpatient telemetry systems are used by hospitals for constant monitoring for serious cardiac events, such as arrhythmia or acute myocardial infarction. Data are transmitted to both the monitors in the patient’s room and the central nursing station. Alarms alert the staff if a serious event is detected.
Outpatient telemetry systems are also available. However, these outpatient telemetry systems are more of an alternative to Holter or event monitoring than a replacement for inpatient telemetry. Patients with very transient and minimally symptomatic arrhythmic episodes (namely atrial fibrillation) are ideal candidates for outpatient telemetry. Similar to an inpatient monitoring system, the ECG is recorded by a transmitter worn by the patient that continuously transmits data to a monitoring station in the home. If the monitoring station detects an arrhythmia, the station will transmit the ECG by landline or cell phone to a center where a technician can overread and confirm the arrhythmia. Doctors can use this information to confirm the presence of arrhythmia and adjust medication on an outpatient basis if needed. Because the captured data are transmitted and not stored, monitoring for indefinite periods of time is possible with only battery changes/recharging needed.

3.11 CARDIAC STRESS TESTING

Cardiac stress testing involves recording an ECG while the patient follows an exercise protocol on a treadmill or stationary bicycle. Infusion of pharmacologic agents can also be used to simulate exercise conditions during a stress test. A stress test can indicate if there is insufficient blood flow to the heart during exertion, which is evident with depression of the ST segment of the ECG. There are special ECG systems designed to work in the stress test laboratory that interface with exercise equipment. Ultrasound or nuclear imaging may also be performed as part of a stress test to better assess heart function and blood flow to the heart muscle.

Specific exercise protocols can be programmed into the stress test systems to control the speed and slope of the treadmill or the resistance for the bicycle test. The Bruce protocol is the most commonly used exercise protocol that is used to screen for coronary artery disease. This protocol consists of seven stages that last 3 min each. In the first stage, the patient begins by walking at speed of 1.7 mph on a 10% incline. In the subsequent stages, the treadmill speed and incline grade are gradually increased until the speed is 6.0 mph and the incline is at 22%. In addition to ECG recording, blood pressure is measured at the end of every stage. A modified Bruce approach is used for patients who are more limited physically. In the modified approach, incline grade begins at 0% instead of 10%.

An example heart rate plot from a patient during a stress test is shown in Figure 3.20. The heart rate sharply increases at the start of exercise and continues to gradually increase.

![Heart Rate Plot](image_url)
before reaching a plateau. When exercise is stopped, heart rate quickly recovers. Both the
ability to reach the target heart rate during exercise (220 – age) and the speed of recovery
once stopping exercise are indicators of cardiovascular health.

3.12 CARDIAC MAPPING
The surface ECG provides composite information of the heart’s electrical activity in its
entirety. However, this “global” view of the electrical activity may not offer sufficient
detail of the activation sequences of the heart to pinpoint the locations responsible
for certain arrhythmias. This detail is particularly needed for radio-frequency ablation
procedures, where catheters are used to deliver radio-frequency energy to specific areas
of the heart to destroy the culprit tissue. Local electrical activity can be recorded by elec-
trodes that are in contact directly with the heart tissue. These contact electrical recordings
are known as electrograms. In a cardiac electrophysiologic study, catheters with electrodes
at the tip are guided through a vein or artery to the heart. X-ray machines that produce
real-time images displayed on monitors are used to guide the catheters. The catheter used
for radio-frequency ablation can also be used to record electrograms. A mapping system is
used to amplify, filter, display, and record electrogram signals. Mapping systems also allow
the selection of channels to perform pacing via an electrical stimulator.

3.12.1 Electrograms
Contact electrograms again differ from surface ECG recordings as they reflect local electri-
cal activity at the site of recording. Electrograms are most commonly either unipolar or
bipolar recordings. Unipolar electrograms are differential recordings made from a single
electrode in contact with the heart referenced to a distant electrode not in contact with the
heart. Figure 3.21 shows the typical waveform of a unipolar electrogram and how it relates
to an activation wavefront. A sharp positive deflection occurs as the electrical propagation

![Unipolar Electrogram Diagram]

FIGURE 3.21 Illustration of the unipolar electrogram and how it relates to activation of tissue.
nears the electrode followed by a sharp negative deflection as the wavefront crosses the electrode and propagates away from the electrode. A slower deflection may follow that corresponds to the repolarization wavefront. The time of maximum negative slope is commonly regarded as the activation time of the recording site. With unipolar electrodes from multiple sites in the heart, the activation sequences can be determined. A greater number of electrodes will produce a more detailed activation map. However, if the cardiac rhythm consists of repetitive activation sequences, the detailed activation maps can be created with one roving catheter that sequentially determines activation times relative to the activation times of one stable electrode.

The morphology of the unipolar electrogram can also provide insight into the activation sequence. If the electrogram only has a negative deflection and not a positive deflection, this can be interpreted as the electrode site being the focal source of an activation sequence. Similarly, if the electrogram only has a positive deflection and no negative deflections, this would indicate that the recording site is at the end of the activation sequence.

A limitation of unipolar electrograms is that far-field activity may also be reflected in the signal, in addition to local activity. The additional of far-field activity may interfere with the ability to determine the activation time from the unipolar electrogram. For this reason, bipolar electrograms are commonly used. Bipolar electrograms are electrical recordings comprised of the difference between two closely spaced electrodes that are both in contact with the heart. Bipolar electrograms can also be thought of as the difference between two unipolar electrograms. The biggest difference between two closely spaced unipolar signals occurs at the time of activation of the two recording locations. Far-field activity, on the other hand, will have similar effects on the closely spaced electrodes and thus will cancel out with the bipolar recording.

The morphology of bipolar electrograms depends on a number of factors. First, unlike unipolar electrograms, bipolar electrograms are dependent on the direction of the activation wavefront. As illustrated in Figure 3.22, a wavefront that first crosses the negative electrode and then crosses the positive electrode will produce a bipolar electrogram that has the opposite polarity to the bipolar electrogram where the wavefront first crosses the positive electrode and then the negative electrode. A wavefront that crosses the positive and negative electrode perfectly perpendicular to the two electrodes will produce a bipolar electrogram with negligible potential, since the two electrodes will sense similar activity. Wider electrode spacing or slower conduction velocity will result in a bipolar deflection with longer duration. High-pass filtering (typically with a 30 Hz cutoff) is typically used to further accentuate the activation deflections during bipolar recordings. With atrial fibrillation, bipolar recording can also appear very fractionated, indicating complex conduction patterns in those areas. These fractionated electrograms have been studied extensively to understand their role in the maintenance of the arrhythmia.

The third kind of electrode recording is known as the monophasic action potential (MAP) recording. A MAP recording is obtained by an electrode that is in contact with the tissue referenced to the second electrode that is in close proximity but not in contact with the tissue. The result is a signal that approximates the shape of the transmembrane action potential. This recording technique is useful in the study of repolarization characteristics...
of the tissue, particularly repolarization dynamics. However, good-quality recordings require the contact electrode to have a constant pressure against the tissue. Suction or spring mechanisms are usually incorporated into the MAP catheter or probe to provide the constant pressure. Despite this, MAP signals will still degrade over a short time. Because of the technical difficulties in achieving good-quality signals, MAPs are rarely obtained during clinical procedures.

### 3.12.2 Effective Refractory Periods

Information of the refractory periods of cardiac tissue is not easily or reliably obtained with unipolar or bipolar recordings. The refractory period can instead be measured using programmed stimulation to pace the tissue. Because refractory periods are dependent on the heart rates, the stimulator is first set to pace at a stable heart rate. A common setting for this phase, known as the S1 interval, is 8 paced beats at a 400 ms interval. Following the S1 intervals, a stimulus with a shorter coupling interval (S2) is delivered. If the S2 stimulus elicits a response from the tissue, then the S1–S2 sequence is repeated with a decremented S2 until the tissue does respond to the S2. The longest S2 interval that fails to exhibit a response is the effective refractory period (ERP) for that particular S1 interval. An illustration of the process of determining the ERP with pacing is shown in Figure 3.23.

### 3.12.3 Electroanatomic Mapping

Determining the catheter locations with fluoroscopy requires a level of intuition by the cardiac electrophysiologist as the heart is not visible from the x-ray images. Furthermore,
the x-ray images are 2D, whereas the catheters are manipulated in 3D space. With complex ablation procedures such as atrial fibrillation ablation that require ablation of many sites, it is important to know what part of the atria has already been ablated. This is not easily determined with a fluoroscopy system alone. An electroanatomic mapping system is a tool utilized in the electrophysiology laboratory to enable tracking of the 3D location of the catheters.

Catheter tracking with electroanatomic mapping works similarly to satellite-based global positioning systems used to track a user’s position on earth. The Carto electroanatomic mapping system by Biosense Webster (Diamond Bar, CA) determines the position of the catheters by triangulating low-level magnetic fields produced by a pad beneath the patient on the table. The Ensite NavX by St. Jude Medical (St. Paul, MN) uses high-frequency currents emitted by patches applied to the skin to triangulate catheter positions. Once the catheters are positioned within the heart chamber of interest, the electroanatomic mapping system will allow real-time tracking of the catheter and reduce the amount of x-ray exposure required for fluoroscopy. By collecting spatial points when the catheter tip is in contact with different parts of the chamber, a 3D geometry corresponding to the geometry of the endocardium can be created. This geometry can be merged with 3D data obtained from MRI or CT for more detailed anatomic information. Electroanatomic mapping can then be used to construct activation maps to determine activation sequences, create voltage

![Figure 3.23 Illustration of pacing-based determination of the ERP. The basic cycle length (S1) is 400 ms. The extrastimulus (S2) is decremented from 200 ms, where an electrogram potential is invoked, to 180 ms, where no electrogram is invoked. The ERP is 180 ms.](image-url)
maps to identify areas of scar, and document ablation sites. Examples of activation maps created by electroanatomic mapping systems are shown in Figure 3.24.

3.12.4 Noncontact Mapping

Contact mapping with catheters is limited by the number of catheters and electrodes available to perform electrogram recording. The St. Jude Medical EnSite system is a noncontact mapping system that records far-field potentials within a chamber of the heart with a multielectrode balloon catheter and uses these signals to predict what the potentials would be on the endocardial surface of the chamber. This process first requires the construction of the 3D geometry of the chamber with electroanatomic mapping and a roving catheter. Each of the 64 electrodes of the balloon catheter records far-field potentials from the entire chamber. By taking account of the distance of each electrode from a point on the wall, a virtual electrogram at that point can be calculated from the inverse solution to Laplace’s law. An unlimited number of virtual electrograms can therefore be computed. A limitation of this system is that areas at greater distance from the balloon will have greater errors in the reconstruction, particularly if there are inaccuracies in the constructed geometry.

3.13 RADIO-FREQUENCY LESION MEASUREMENTS

3.13.1 Contact Sensing and Force Measurement

Electrogram recording and ablation require good contact of the catheter electrodes with the heart. Adequate contact force is a requirement for ability to make transmural ablation lesions. Lesions that are incomplete transmurally may not block activations as intended. Too much contact force comes with the risk of perforation. Thus, contact sensing is important.
to provide an electrophysiologist feedback regarding the amount of pressure the catheter
has against the chamber wall, which is not easily determined from tactile feedback.

The two most common means to determine contact are electrogram signal quality and
impedance measurement. Good-quality electrograms with high amplitude are possible
only with good contact with the chamber wall. Impedance measurements drop significantly
when the catheter is in contact with the heart. However, both these indications only indi-
crectly measure contact force, and both have significant limitations. Electrogram amplitude
by itself is not a good indicator of contact as both poor contact or scar tissue can be respon-
sible for low amplitude. Neither electrogram amplitude nor impedance can provide good
indication that too much force is being applied. Thus, more direct measures of contact
force would be desirable.

There are a few technologies that are being developed and evaluated for contact force
measurement. A catheter by Endosense (Geneva, Switzerland) uses optical fibers to
sense catheter contact force. Bending of the optical fibers produces a change in wave-
length that can be translated into a force measurement. A second technology by Biosense
Webster (Diamond Bar, CA) uses a spring-loaded tip with a magnetic signal emitter and
three magnetic sensors. Pressure applied to the tip will change the position of the sen-
sors relative to the emitter and results in the change of the detected signal. The contact
force is estimated based on the signal change. For both technologies, force is measured
in grams. Because of the movement associated with the beating heart, contact force will
vary during the time needed to apply the radio-frequency energy. To determine the
total amount of contact “energy,” a force–time integral can be calculated. The measured
force–time integral has been shown to be a better predictor of lesion size than the peak
force of the catheter.

3.13.2 Tissue Temperature Sensing

Radio-frequency ablation destroys tissue by heating the site of catheter contact. The radio-
frequency current density is the highest at the point of contact; thus, most of the heat will
be generated there, as current through a resistive medium generates heat. The temperature
to which the tissue is heated is a factor in the formation of the lesion. Thus, temperature
sensors on ablation catheters are useful in determining the power output needed to create
adequate lesions to also prevent overheating. Thermistors and thermocouples are two com-
mon types of temperature sensors. Thermistors change resistance in response to changing
temperature. By applying a constant current to the thermistor, a voltage is produced that
decreases as temperature increases. Thermocouples are sensors that consist of two different
conductors, which output a voltage proportional to the temperature. Unlike thermistors,
thermocouples do not need to have current applied to them to output a voltage.

We previously discussed the importance of contact force and contact time for lesion
generation. Convective heat transfer by the circulating blood pool is a counteracting
factor against lesion formation. The greater the convective cooling, the greater amount
of radio-frequency output that is needed to obtain the desired effect. Measurement of
the convective heat transfer coefficient can be performed by measuring the tempera-
ture with two sensors: one at the tip and another nearby, which measures blood flow
temperature. The difference between the temperatures divided by the product of the electric power consumed by heating the sensor and sensor area equals the convective heat transfer coefficient.

3.14 POSSIBLE FUTURE INSTRUMENTS

ECG and contact electrical recordings of the heart have stood the test of time for their utility in the diagnosis of arrhythmias and other heart conditions. The current techniques for ECG measurement may not significantly evolve from their present forms, although the hardware may improve and new algorithms used for automatic diagnosis or risk stratification may be developed. What we may see more of in the future is the integration of ECG with other imaging modalities to provide the context of heart function and substrate with the electrical activity. Three-dimensional MRI or CT can provide detailed characterization of the torso, the heart, and even scarring within the heart. How to use this information in conjunction with ECG body surface mapping to accurately characterize activation sequences in the heart is a subject of active investigation. If local conduction velocities and repolarization can be measured in this fashion, patient-specific computer simulation could be used to test arrhythmia vulnerability in the modeled heart and perhaps be used to plan ablation or other course of treatment. This would be a noninvasive and lower-risk alternative to a study performed in the cardiac electrophysiology laboratory involving catheter insertion and x-ray exposure.

The greatest number of sudden cardiac deaths occurs in the population with no known risk factors. Future devices that are low cost and minimally intrusive to monitor ECG could provide alerts to emergency services when arrhythmias are detected and could potentially save many lives. It would need to be determined whether this technology could be practically integrated into cell phones, watches, personal digital assistants, or made as a subcutaneous device.

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