Signals and Signal Processing for the Electrophysiologist
Part II: Signal Processing and Artifact

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In part II of this series, using the background covered in part I on electrogram acquisition and signal processing, we discuss potential errors in mapping because of artifact or inappropriate filtering of signals that may lead to unsuitable selection of ablation sites.

Artifact

Saturation Artifact
If a 3-mV peak signal is being amplified at a gain of 5000, the final output will be 15 000 mV or 15 V; however, if the system supply voltage is limited to 10 V, the overall signal amplitude would be limited to about 9.5 V (slightly below the supply voltage). The signal would simply be cut off above 9.5 V, although the amplifiers would attempt to drive themselves up to the required 15 V output unsuccessfully. The amplifiers remain “saturated” during this time, and the top of the signal flattens out and remains flattened until the signal amplitude decreases to below 9.5 V, when it can be tracked with fidelity once again. This is a form of distortion and can only be avoided by reducing the gain of the system (in this numeric example, to less than 3000) (Figure 1). When amplifiers saturate, they can take several milliseconds to recover, further delaying the time to faithful reproduction of signals. This is exacerbated with extremely large signals, such as pacing signals, which are typically 3 orders of magnitude larger than the electrograms. In this case, the signal recovery time for the paced channel as well as adjacent electrodes can be long enough that the artifact envelops a substantial portion of electrogram information immediately following the pacing spike (Figure 2). This can be a troublesome problem when attempting to measure postspacing intervals after entraining a tachycardia. Despite this artifact, the operator must make every effort to use the paced electrode signals for this measurement, as using more distant electrodes may produce misleading results. Saturation from pacing creates an artifactual difficulty in measuring the postspacing interval on the mapping catheter.

Artifact Because of Clipping
Another disadvantage of using high gain settings is that multiple signals overlap on the display, making analysis difficult. This can be minimized by artificially “clipping” the signals of interest. The advantage of clipping (usually done at the software stage, although it can be done in hardware also) is that it can limit the amplitude of a displayed signal (in volts) numerically (in the mathematical space of computer software), without suffering the ill effects of hardware distortion and signal recovery times associated with saturation of the amplifiers (Figure 1). This would allow the electrophysiologist to see a small electrogram that immediately follows a very large clipped electrogram without overlapping the displayed signals on adjacent channels; however, the significant disadvantage of clipping is that one could get easily misled when looking at fused waveforms (pathway potentials during an accessory pathway ablation, for instance) if the selected clipping level by the software prevents visualization of 2 discrete, fused signals and instead represents them as a single, wide signal (Figure 3). The operator cannot control the clipping level as this is programmed into the software in most systems. In this setting, one is almost always better off deliberately running the amplifiers at a low gain in order to prevent amplifier saturation and not require clipping (to avoid adjacent channel signal overlap). The only disadvantage with this approach is that great care has to be taken to look at every nuance of the signals to not miss small potentials, such as pulmonary vein potentials or accessory pathway potentials, which are difficult to see at low gain settings, especially when the operator is several feet away from the display screen. In general, it’s better to avoid clipping signals but rather to gain them down and analyze each component specifically.

Phantom Signals
Phantom signals (signals that appear real but are artifact) can also be seen because of the less-than-perfect common-mode rejection ratio of the input amplifiers, especially at higher frequencies. If the amplifiers were ideal, they would be able to reject any coupled interference as long as it was equally
coupled to both inputs; however, at high gain settings, even if a catheter is only partially in a sheath or is not even in the body, unequally coupled electric interference from pacing spikes (which have substantial high-frequency energy and are easily capacitively coupled to nearby electrodes) can mimic real signals at the pacing rate (Figure 4A), especially after filtering smoothes out the interference and makes it look like a real far-field signal (Figure 4B). This is because the catheter itself acts like an antenna and is connected to the inputs of its own amplifier, amplifying any information provided to its inputs. Programming the output currents of the pacing stimulator to no more than twice the diastolic threshold (to minimize electric and magnetic coupling) at a fairly narrow pulse width of 1 to 2 ms (to reduce overall energy), confirming that all catheters are well-extended out of the sheath tips, and making good contact between tissue and tip all help to minimize this problem, both by reducing interference and increasing the required signal amplitude. Interference may also be minimized on an ablation catheter tip by adding a second return patch (from the patient’s skin to the radiofrequency [RF] generator) since this also lowers effective contact impedance and reduces coupled interference.

Programming pacing output currents to lowest possible energy and using a second return patch to the patient’s skin minimizes phantom signals.

Coupled Noise Because of Radiofrequency Ablation

Delivering 50 W into a 100-Ohm impedance requires 70 Vrms (200 V peak-peak sine wave). This high-amplitude ablation signal needs to be filtered and rejected effectively so it does not obscure the required intracardiac electrograms that are 200 000 times smaller. As discussed earlier, hardware manufacturers use sophisticated circuitry to help filter and reject the RF energy and only display the low-frequency intracardiac information; however, in addition to the instrumentation amplifiers and filters, the internal cabling of the catheters is extremely important. Sensitive pairs of electrodes usually have twisted-pair wiring in their cabling, with improved performance obtained by increasing the number of twists-per-inch. This is because coupled interference (from a remote source) is more likely to be picked up equally by both wires if they are very close to each other. Increasing the twists-per-inch increases the proximity of the 2 wires and the chance of equal pickup, allowing the input amplifiers to reject the interference more effectively. Any imbalance in the pickup will result in large amounts of interference. Whenever a channel is enabled for pacing, a constant-current source is attached to the electrode inputs in preparation for pacing. This unbalances the impedances seen at the 2 electrode inputs, reducing the efficacy of rejection by the input instrumentation.
amplifiers. As a result, coupled interference in the form of high-frequency noise and large baseline drifts (because of rectification of the high frequencies by the semiconductor junctions in the electronic circuitry) will be superimposed on the required electrograms, making observation of reduction in electrogram amplitude difficult or even impossible. Turning off the pacing input usually takes care of this problem, especially on the mapping catheters that are close to the ablation location (Figure 5), although it can take several seconds for full recovery. The pacing channel input should be turned off when ablating or continuing mapping.

Also, the amount of noise coupled on to a particular electrode pair depends on the impedance between that electrode pair. This is because the internal impedance of the source of interference and the interelectrode impedance act as a potential divider. The greater the coupling impedance (internal impedance of the coupling source), and the lower the interelectrode impedance, the less the coupled interference. This is essentially like turning down the volume control on an audio player, which increases the effective source impedance and reduces the effective load impedance, lowering the signal amplitude and thereby the sound volume.

There is, however, a limit to the amount of interference rejection that can be achieved this way on the ablation catheter because the ablation energy is usually delivered only between the distal electrode and the return patch on the patient’s skin, producing an intrinsic imbalance in coupling. Additional circuitry in the patient module (such as notch filtering at the ablation frequency) and passive filters that increase rejection at the high frequencies, without the limitations faced by active “amplifier-based” filters, usually takes care of the problem. This artifact may also be affected by the difference in size between the tip electrode and second electrode (the pair usually connected to the mapping amplifier in bipolar mode). In the extreme case of an 8-mm ablation tip, for example, the ablation energy is being delivered to the

Figure 3. A. Fusion of signals during clipping cause obscuration of retrograde accessory pathway potential on the ablation catheter during last paced beat. B. Also shown is the “smearing” of double potentials on the ABL_r in another patient with clipping activated, almost obscuring the second potential. The left-hand panels in both A and B show the unclipped signals, with the respective clipped signals shown in the right-hand panels.

Figure 4. A. Phantom interference (pickup) by the His1 channel during pacing from the right ventricular (RV) catheter. The stimulator is attempting to pace the RV at a cycle length of 500 ms, but the pacing function has not been enabled on the data acquisition system. Pacing spikes are capacitively coupled to the His1 channel and manifest as small potentials not synchronous with the intrinsic rhythm (arrows). B. Phantom pickup on the ABL_d (catheter in left atrium) with an unconnected ABL_r. The unconnected ABL_r amplifier input picks up a far-field ventricular signal, probably from adjacent signals on the connector block. The ABL_d signal (proximal electrode pair on the same catheter) is partially within the long sheath and shows increased noise, although the pickup is slightly visible faintly.
distal electrode, which is presumably in contact with the tissue, with the RF energy being coupled in variable amounts to the second electrode (usually only 2-mm wide), which is not necessarily making tissue contact. This can produce a significant impedance imbalance from each electrode to circuit ground, again affecting input amplifier balance and coupled interference. Although the distance between the distal and No. 2 electrodes may be the same in 8-mm or 5-mm tip electrodes, the coupled external interference between the 8-mm tip and the second electrode will be larger than the coupled external interference between the 5-mm tip and the second electrode (because of the greater difference in surface areas between the 2 electrodes, which translates to a larger difference in impedance between the 2 electrodes and thereby a greater difference in coupled interference, which leads to greater noise). Also, higher ablation energies are used with the larger tip electrodes, also increasing likelihood of coupled interference from the RF source itself. Using a smaller tip ablation catheter can minimize this imbalance and interference at the expense of a possibly less-effective ablation lesion. Another reasonable compromise may be to use an externally irrigated catheter, which usually has a small mapping tip (3.5 mm), providing better balance between the 2 bipolar electrodes; however, the very act of irrigating externally with saline may change the effective distal electrode area (making it bigger, depending on the rate of irrigation). As long as the catheter tip maintains tissue contact, its overall impedance to the amplifier reference does not change significantly. This is because the impedance to the surrounding tissue attributable to coupling by the saline irrigation shunts the impedance between the tip and amplifier reference, and the parallel combination is not appreciably smaller than the tip impedance itself (as measured by the authors in an animal model). The electrogram signal amplitude may still be affected since this is determined by the overall coupled signal to the distal electrode with respect to the proximal electrode (second electrode). If some of the irrigation holes are blocked because of the tip not being perpendicular to the tissue, the irrigated area changes, affecting both the ablation lesion as well as coupled electrogram. So a change in tip orientation could reduce electrogram size and mislead the operator into thinking that an effective ablation lesion has been provided. This could be less of an issue with nonirrigated or internally irrigated catheters.

Electrode Polarization Artifact
The silver-silver chloride ECG electrodes, as well as the metal alloy catheter electrodes, can develop small voltages on their surfaces as an electrophysiologic (EP) study proceeds because of small ionic currents traveling between the various electrodes in a conductive medium. Although most amplifiers have high pass filters (with the possible exception of unipolar amplifiers) and will block this DC voltage, the slowly changing DC levels do get transmitted to the output, resulting in slowly drifting baselines. This drift can temporarily obscure small signals and can impede accurate mapping or localization. This will be more obvious on the ECG channels or unipolar channels with a low-frequency cutoff of 0.05 Hz. Replacing ECG pads, removing excess moisture on the patient’s skin, and better skin preparation before ECG pad placement will help substantially.
Motion Artifact
Rhythmic movement (because of cardiac motion, ventricular assist devices, infusion pumps, or forced-air warming blankets) can produce misleading signals (Figure 6), which can resemble far-field intracardiac signals or even fractionated potentials from a scar. This may be suspected if catheter repositioning continues to produce the same signal, suggesting an external source. Sequentially eliminating each possible source usually isolates the problem.

Electrode Contact Artifact
Intermittent contact between electrodes on different catheters or electrodes on the same catheter (such as a variable-diameter lasso catheter with overlapping electrodes or contact between an ablation catheter and mapping catheter) can produce misleading signals. In the case of a lasso catheter, one could be misled into deducing a pulmonary vein exit block because of noise on several overlapping electrodes that would produce sharp, near-field signals reminiscent of pulmonary vein potentials, which do not get transmitted to the atrium. This may also be seen with a fixed-diameter lasso, which has been pushed into a pulmonary vein ostium with consequent overlap of electrodes. Correlation of the location of these “potentials” with fluoroscopic electrode information reveals the true nature of these signals (Figure 7).

Bipolar and Unipolar Signals
In reality, all recorded signals in the EP laboratory are bipolar in that they constitute a potential difference between 2 electrodes. These 2 electrodes are connected to the 2 inputs of an instrumentation amplifier (which has been described in detail in the overview section previously); however, when 1 of the electrodes is some distance from the cardiac structures (ideally at infinity), the potential difference between the exploring electrode (a term originally used by Frank Wilson in 1932) and this distantly positioned electrode (indifferent electrode) is referred to as a unipolar signal. On the 12-lead ECG, precordial leads V1 through V6 record voltages between each of the precordial wires and the indifferent electrode, which is the Wilson Central Terminal (WCT). WCT is
assumed to be at the center of a triangle with the 3 limb wires at the apices (Figure 8). The vector sum of currents entering the WCT is theoretically 0, and this is considered to be a simplified electric center of the heart. In early studies by Wilson,3a Frank,3b Goldberger,3c and others, attempts were made to either immerse the patient in a saline bath and use the surrounding saline as the indifferent electrode or use a ground patch on the patient as an indifferent electrode. For reasons of convenience as well as a better ability to reject coupled interference on the skin (by actually applying a voltage to the patient’s skin via the right leg driver [see discussion in Part 1 of this series]), the WCT concept continues to be employed. When intracardiac electrograms need to be displayed in unipolar fashion, the WCT may be used as the reference, with the exploring electrode being the distal electrode of a mapping catheter. The mapping system can be configured to display these unipolar signals directly. Electroanatomic mapping systems generate their own WCT and display an independent unipolar electrogram on their dedicated displays. Alternatively, a catheter may be placed in the inferior vena cava, and this catheter’s electrodes may be used as the indifferent electrode for unipolar signals. Some centers use custom catheters with extra electrodes positioned proximally in the inferior vena cava with the distal electrodes at the His bundle location. The inferior vena cava electrodes may serve as indifferent electrodes for unipolar recording or pacing, as needed.4

Using the same ECG rules for vectors and polarity, a wavefront of depolarization that approaches a unipolar exploring electrode generates a positive deflection on the display. As this wavefront continues past the electrode, a negative deflection is displayed. The transition from positive to negative occurs when the wavefront is immediately beneath the exploring electrode and is marked by the steepest negative slope of the signal. This concept is only valid when the low-frequency cutoff for the amplifier channel is kept suitably low (DC or 0.5 Hz maximum instead of the usual 30 Hz cutoff used for bipolar channels). The instantaneous amplitude of the voltage is directly proportional to the area of the wavefront of depolarization and inversely proportional to the square of its distance from the exploring electrode.5,6

Since the signal is theoretically approaching or receding from the unipolar electrode with respect to infinity, no specific directionality (lateral, medial, superior, inferior) is obtained from unipolar recordings.

If the wavefront of depolarization originates immediately below the exploring electrode, it spreads away from the exploring electrode in all directions simultaneously, producing a negative deflection with no R wave associated with it. This is usually a site of successful ablation of an automatic

![Figure 8. Derived Wilson Central Terminal (WCT) (from ECG limb leads) used as “indifferent” electrode for ECG unipolar leads (V1 to V6) as well as for unipolar catheter mapping. Any noise or baseline drift on the limb electrodes RA, LA, and LL will cause the WCT voltage to be noisy, affecting the precordial lead signals as well as the unipolar signal on a mapping or ablation catheter. The 12-lead ECG shows an increase in baseline drift and noise when the Right Leg Driver is disconnected from patient’s body (left panel with right leg driver connected and right panel with right leg driver disconnected).](http://circep.ahajournals.org/doi/abs/10.1161/CIRCEP.117.004888)
tachycardia. Having a very small R wave associated with this predominantly negative deflection does not necessarily rule out this location as the origin since the focus could be intramyocardial at the same location; however, ablating in this location does not always produce success because of the relatively large area over which a negative deflection may be seen.7 If the filter settings for the unipolar channel are maintained from 30 Hz to 500 Hz, a large portion of the unipolar signal will be modified, causing artificial R waves to appear on the signal, even if the catheter is at the origin of the impulse. Maintaining the high-frequency cutoff at 500 Hz but reducing the low frequency cutoff to 0.5 Hz or even DC (if possible) will preserve nearly all of the information in the unipolar signal. Poor electrode contact (or no electrode contact, as is seen if the electrode is floating in the cardiac chamber) manifests as a slow (low rate-of-change) signal instead of the expected rapid positive-to-negative polarity change. Using the unipolar signal to locate accessory pathways and ablate successfully has also been described, with the best results being obtained with a P-QS pattern (the atrial signal fused to the unipolar QS signal) on an “unfiltered” (no high-pass filtering) signal.8

Disadvantages of Unipolar Electrodes

Even though, by the inverse square law, electrogram amplitude helps differentiate near-field from far-field signals on the unipolar electrogram, one can be misled in the setting of mapping diseased tissue with low-amplitude, near-field information when comparing it with high-amplitude, far-field information from distant, healthy tissue. The relatively large contribution of the healthy, distant tissues can obscure the small, local contribution of the scarred myocardium. This effect is exacerbated by the low cutoff frequencies usually recommended for the high-pass filter. So, in this setting, using a higher cutoff frequency such as 10 Hz or even 30 Hz helps minimize the far-field contribution without completely losing information about the approach or recession of signals.4

Another disadvantage of unipolar signals is the increased susceptibility to noise because of the differential coupling of interfering signals to the 2 widely spaced electrodes. This can be particularly irksome when using the WCT as the indifferent electrode, because of surface electrode polarization and contact noise. Using an internal, indifferent electrode in the inferior vena cava instead may ameliorate this problem.

Bipolar Electrograms

When 2 narrowly spaced electrodes (less than 2 to 3 mm apart) in contact with myocardium are attached to the inputs of an instrumentation amplifier, the difference between the 2 signals constitutes a bipolar electrogram, the most common waveform analyzed on the display during a typical EP study. Successive pairs of closely spaced electrodes on a multipolar catheter (1 to 2, 3 to 4, 5 to 6, etc) or even contiguous adjacent pairs (1 to 2, 2 to 3, 3 to 4, etc) produce reasonable bipolar signals, as long as the electrode spacing is small. The same bipolar information may be obtained by subtracting 1 unipolar signal from another unipolar signal, as long as the 2 unipolar electrodes are close to each other and in contact with myocardium. These signals are a good indicator of local electric activity immediately beneath the 2 closely spaced electrodes and so can provide information on timing of local activation.9 Local activation is denoted by the time of maximum deflection of the electrogram. Reducing the distance between the 2 electrodes of the bipole can further reduce the sensitivity to distant activation events.10 The amplitude of these signals in healthy ventricular myocardium (measured with a 10-mm electrode spacing) ranges from 3 mV to 10 mV with durations of less than 70 m.11 These amplitudes are reduced substantially with smaller spacing. Also, a lot of the far-field information present in a unipolar signal is removed from the bipolar signal because each electrode of the bipolar pair “sees” a similar (or identical) far-field voltage, and these similar signals are canceled out by the instrumentation amplifier that subtracts 1 electrode’s voltage from the other electrode’s voltage. Thus, it is reasonable to use a high-pass filter with a higher cutoff frequency, such as 30 Hz, for bipolar signals, since this will help minimize baseline drifts/shifts, with no further loss of information.

With close electrode spacing, changes in polarity on the bipolar signals correlate with changes in direction of activation, which may occur during creation of a conduction block during RF ablation. In the extreme case, this could be a disadvantage since a wave of depolarization that is perpendicular to the 2 electrodes would be canceled.10 Conversely, a wave of depolarization that is parallel to the 2 electrodes would be picked up sequentially by both, resulting in maximum amplitude. The time of peak amplitude in a bipolar signal is the equivalent of the fastest negative downslope of the unipolar signal. Furthermore, if both minimally filtered unipolar signals and standard-filtered (30 to 500 Hz) bipolar signals are obtained simultaneously, the time between onset of the unipolar electrogram and first peak of the bipolar electrogram is usually very short (less than 15 m) at the site of successful ablation of an automatic tachycardia.12

Given the difference in size between the tip electrode and the more proximal electrode of a bipole pair, and the fact that, in most cases, the tip is in contact with the tissue with perpendicular orientation, leaving the proximal electrode offset from the tissue by several millimeters, it is not clear if the voltage on the bipolar electrogram is truly representative of the “quality of tissue” beneath the catheter tip. This uncertainty becomes particularly significant when creating voltage maps of ventricular myocardium and using absolute amplitude cutoffs to differentiate a scar from viable tissue. This is of somewhat lower concern during epicardial mapping since both electrodes are usually making contact with the tissue because of the more parallel orientation of the catheter “trapped” between epicardium and pericardium. During epicardial mapping, however, areas with more than 5 mm of epicardial fat thickness (near the base of the heart) tend to have lower bipolar voltages (less than 1.5 mV, possibly being misclassified as a scar), compared to other areas.13

One can be misled by interpreting electrograms obtained from more widely spaced electrodes (for example the 5-mm spacing on a lasso-style catheter used in atrial fibrillation ablation) as true bipolar electrograms. Because of the wide spacing, the difference signal obtained between the 2 ele-
trodes does not inform us about the tissue potentials immediately below the electrodes (since these could be widely different because of the spacing). Attempting to “time” such a signal by looking at the peak amplitude would be incorrect. This can be corrected by using the ablation catheter to map in the general region of this early “bipole” and finding a location that is early on the ablation catheter before delivering energy.

Another subtle but important point is that bipolar electrograms on the distal pair of electrodes of the ablation catheter do not always decrease during ablation, the usual indicator of ablation success. Since the amplifier is calculating the difference between the distal electrode and proximal electrode potentials to generate the electrogram, as the tissue below the distal electrode is ablated, this difference could decrease (as we anticipate) or even increase (with reverse polarity), since the proximal electrode may have a larger signal now because of its sampling of more healthy, unablated tissue compared with the desiccated tissue beneath the distal electrode. A true unipolar electrogram of the distal electrode would, however, unequivocally decrease as the tissue is ablated effectively.

Conclusion

Acquiring and interpreting intracardiac electrograms in the noisy, unpredictable environment of the modern EP laboratory is fraught with difficulties; however, a combination of good hardware and software design, careful management of noise sources and cabling and reasoned interpretation of the presented data allow considerable success during mapping and ablation of cardiac arrhythmias.

Disclosures

None.

References


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