Top-Resolution Frequency Analysis of Electrocardiogram With Adaptive Frequency Determination

Identification of Late Potentials in Patients With Coronary Artery Disease

Ralph Haberl, MD, Hans F. Schels, Peter Steinbigler, MD, Gerhard Jilge, and Gerhard Steinbeck, MD

Frequency analysis of the electrocardiogram with Fourier transform is a sensitive method of detecting late potentials. However, information about localization of late potentials is lost, frequency resolution is poor, and window functions have to be applied. We therefore analyzed multiple segments (25 msec long) of the surface electrocardiogram (“spectrotemporal mapping”) with adaptive frequency determination (AFD), an autoregressive algorithm that is characterized by high-frequency resolution in very short segments without the use of window functions. Results were compared with those from Fourier transform and the Simson method.

We studied 38 patients after myocardial infarction (MI) with sustained ventricular tachycardia (VT), 21 patients after MI without VT, and 18 healthy subjects. Frequency peaks could be clearly differentiated until a minimal interval of 6 Hz; with fast Fourier transform (Blackman Harris window) in a much longer segment (80 msec), the spectral peaks merged into one another at an interval of about 30 Hz. AFD revealed high-frequency components as narrow peaks in the range of 40-160 Hz in 28 of 38 patients (74%) after MI with VT. Because of the short segment size, exact localization of late potentials was possible; in most of the patients, the peaks occurred in segments inside the QRS complex and ended 20±10 msec after termination of the QRS complex. In patients after MI without VT, only four of 21 patients (19%) had spectral peaks in segments after the end of the QRS complex; however, 13 of 21 patients demonstrated microvolt potentials in segments within the QRS complex. These potentials did not extend beyond the end of normal ventricular activation. Only two of 18 healthy subjects showed abnormal AFD results. Patients with bundle branch block did not need to be excluded.

AFD allowed good differentiation between late potentials and noise by a characteristic pattern of the spectral peaks. For the Simson method, patients with bundle branch block had to be excluded, and overall sensitivity was 42%. In five cases, the cause of failure of the Simson method could be identified as incorrect determination of the QRS limits due to noise. Thus, AFD is a promising method for detailed analysis of late potentials; it combines the advantages of frequency analysis (good differentiation between signal and noise and high-pass filters not necessary) and time domain analysis (localization of late potentials). (Circulation 1990;82:1183-1192)
TABLE 1. Patient Data

<table>
<thead>
<tr>
<th></th>
<th>Group 1 (MI with VT)</th>
<th>Group 2 (MI without VT)</th>
<th>Group 3 (healthy persons)</th>
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</thead>
<tbody>
<tr>
<td><strong>n</strong></td>
<td>38</td>
<td>21</td>
<td>18</td>
</tr>
<tr>
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<td>18/3</td>
<td>11/7</td>
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<tr>
<td><strong>Age (yr)</strong></td>
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<td>55±11</td>
<td>29±11</td>
</tr>
<tr>
<td><strong>Site of infarction</strong></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
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<td>...</td>
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<tr>
<td>Posterior</td>
<td>15</td>
<td>10</td>
<td>...</td>
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<tr>
<td><strong>Ejection fraction (%)</strong></td>
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<td>49±14</td>
<td>...</td>
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<tr>
<td><strong>Bundle branch block (n)</strong></td>
<td>11</td>
<td>7</td>
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</table>

MI, myocardial infarction; VT, ventricular tachycardia.

so-called “window functions” are necessary, which may attenuate the signal of interest.

Based on encouraging results in techniques, geophysics, and biomedicine,6–9 we developed a new alternative method for computation of the frequency spectrum, the adaptive frequency determination (AFD). This is an autoregressive model that is characterized by high-frequency resolution despite small sample size and very good noise rejection and does not require window functions. This method combines the advantages of time domain analysis (localization of late potentials) and frequency domain analysis (filters not necessary and better discrimination of noise).

Methods

Orthogonal bipolar electrocardiograms were recorded from the body surface with a special low-noise, high-gain amplifier as previously described. 10 Three channels were recorded simultaneously and digitized with an accuracy of 12 bits at a sampling rate of 1,000 Hz. Signal averaging was performed until the root-mean-square noise level during diastole was less than 0.3 μV. Each beat was compared with a template before averaging to eliminate extrasystoles and grossly noisy signals. The end of the QRS complex was evaluated by the computer; the spatial vector velocity was calculated; and the end of QRS was defined by the peak of the autocorrelation function. 11,12

In the time domain, late potentials in single leads were defined as deflections of more than 1 μV (more than threefold the basic noise) after the end of QRS in all three channels. The Simson method was applied as a reference (bidirectional filter at 25 Hz) 13,14; abnormal findings were defined as duration of the filtered QRS complex of 120 msec or more and a root-mean-square voltage of the last 40 msec of the QRS complex of less than 25 μV. Patients with QRS duration of more than 120 msec due to left bundle branch block were excluded for the Simson method but not for frequency analysis [classic left bundle branch block in all patients (Table 1)].

Spectrum analysis of the single leads was performed with AFD derived from the adaptive forward backward least-squares method, an autoregressive model. 15 The basic idea of AFD is to calculate the power spectrum by extrapolation of the autocorrelation function of the time domain signal by calculation of a number of “coefficients” in an autoregressive mode.2,4,16–23 The sophisticated mathematical algorithm (see “Appendix”) acts in the following manner (Figure 1). It extracts information from the time domain signal by the evaluation of coefficients. Each time a new coefficient is added, the response of these coefficients is calculated and compared with the signal. This information is used to update the existing coefficients and calculate the next ones; the process ends as soon as no additional relevant information is found. The algorithm was designed for stationary signals; in this study, the signals are considered stationarilike because they are signal-averaged, the means of the data values were subtracted before frequency analysis, and the subsegments were very short (25 msec) with small variances. The advantages of this method are that 1) the time domain signal does not have to be multiplied with a window function before spectral analysis;2) spectral leakage is low; and 3) the frequency resolution is very high, even in segments as short as 25 msec.

In Figure 2, a test signal (addition of two nonstationary microvolt signals and noise on a slope) is processed with fast Fourier transform (Blackman Harris window) and AFD (no window). AFD detects and differentiates the two components as narrow peaks as soon as at least 4 msec of the waveform is included in the segment (25 msec long). The Fourier transform discriminates just the two components of

Figure 1. Principle of adaptive frequency determination (AFD). A time domain signal (upper panel) such as a test signal or electrocardiogram is processed with AFD. Filter coefficients are calculated by computer in a regressive loop to estimate time domain signal as close as possible. Filter coefficients are basis for calculation of power spectrum (lower panel).
the signal, although the segment size is much longer (80 msec). The spectral peaks are very broad; the high peak at 20 Hz is caused by the slope, and the basic noise level is much higher.

A critical point with autoregressive models is the determination of the order (equals the optimal number of coefficients necessary for adequate estimation of the spectrum). If the number of coefficients is too small, components of a signal may be lost; if the number is too high, artificial peaks may occur (Figure 3). For AFD, no stop criterion is referenced in the literature. Therefore, we developed our own criterion to determine the optimal order of the AFD coefficients; the algorithm starts with order equal to segment size (maximum, 50). After each recursion step, the prediction errors are determined; as soon as the errors are outside a predefined range, the optimal order is found. Then the algorithm starts again

**Figure 2.** Analysis of a test signal with adaptive frequency determination (AFD). Nonstationary microvolt signals (sine waves of 70 and 120 Hz multiplied with an e-function; maximal amplitude, 5 \( \mu \)V) are added to a 60° slope with basic noise of 2 \( \mu \)V (upper panel). Multiple segments are processed with AFD (segment size, 25 msec; no window) and Fourier transform (segment size, 80 msec; Blackman Harris window). With fast Fourier transform, spectral peaks are broad (poor frequency resolution), a high peak at fundamental frequency is caused by slope, basic noise level is high, and frequency components can hardly be differentiated. AFD is characterized by narrow peaks (high-frequency resolution); peak at 20 Hz does not influence evaluation of high-frequency components, basic noise level is lower (see segments 30–50), and frequency components can be clearly separated.

**Figure 3.** Influence of order of autoregressive model. A test signal with two sine waves at 60 and 120 Hz on a 45° ramp was analyzed with order 3, 20, and 28, respectively. If order is too low, 120-Hz component cannot be identified. With optimal order (20 in this example), frequency components at 60 and 120 Hz are correctly evaluated. If order is too high, artificial peaks (indicated with *) may occur. Peak at very low frequency (about 30 Hz) is caused by ramp.
with this optimal order. In addition, the stability test for adaptive filters is used.\textsuperscript{19}

A difficult problem arises with all spectral methods if the signals of interest (i.e., late potentials) are located within a steep slope (i.e., in the ascending ST wave). AFD, however, offers the unique ability to attenuate the fundamental slow components without conventional high-pass filtering, thereby circumventing the well-known disadvantages of such filters. The time segment is analyzed first with AFD of low order (from 1 to 3 describes the slow, high-amplitude components of the ST wave). Then, AFD is repeated on the same signal with optimal order (about 14–20; determined by the computer). The two spectra are subtracted to build a “difference spectrum.” The slow components of the signal (i.e., due to the ascending S wave) are strongly attenuated; this allows a much better evaluation of the high-frequency components (i.e., due to late potentials), which were changed only insignificantly by building of the difference spectrum.

To discriminate between noise and late potentials, 38 segments of the ST wave (25 msec long) were analyzed with AFD (Figure 4), the first of which began 42 msec after the end of the QRS complex. The subsequent segments began progressively earlier in the cardiac cycle in steps of 2 msec; thus, two neighboring segments overlapped by 23 msec. The power spectra (not normalized) were combined into a three-dimensional graph (spectrotemporal mapping).\textsuperscript{10}

Spectral values below the mean diastolic noise level (0.3 $\mu$V) were clipped and set to zero dB (dynamic range, $\pm 60$ dB).

To classify the AFD results, a factor of normality was defined: The absolute spectral area in the range of 40–120 Hz was calculated in segments at the end of the QRS complex (segments 16–28; area 1) and in segments far outside the QRS complex (segments 1–12; area 2). The factor of normality was given by area 2 divided by area 1 and is expressed in percent of the total area. A factor of 30% or less was considered abnormal.

The beginning and the end of the late potentials were determined as the starting points of the spectrum (with respect to the end of QRS), which showed a 10% increase of the spectral area compared with noise level. The actual start of the late potentials is about 4 msec earlier and the actual end is 4 msec later in the cardiac cycle because a minimal period of 4 msec of late potentials must be within a segment to produce a spectral peak (3–5 msec depending on waveform and frequency content). The data in Table 2 are already corrected for by this factor.

As a reference method, frequency analysis of multiple segments with fast Fourier transform was also performed in each patient as described previously.\textsuperscript{2,10} With fast Fourier transform, the factor of normality was calculated by comparison of the spectra with cross-correlation\textsuperscript{2} (see “Discussion”). Again, a factor of normality of 30% or less was considered abnormal.

Three groups of patients were studied. Group 1 comprised 38 patients after myocardial infarction (median time after infarction, 26 months; range, 6–34 months) with documented spontaneous sustained ventricular tachycardia. In 34 of 38 patients, the tachycardia could be initiated by programmed ventricular stimulation (one or two extrastimuli at basic rates of 100, 120, 150, or 180 beats/min; stimulus duration, 2 msec; twice-diastolic threshold; stimulation sites: right ventricular apex and septum). A tachycardia was considered sustained if it lasted more than 30 seconds or required therapeutic interventions because of hemodynamic deterioration. In the remaining four patients, a nonsustained ventricular tachycardia (5–12 beats) was induced.

Group 2 comprised 21 patients after myocardial infarction (median time after infarction, 27 months; range, 5–36 months) with no clinical signs of arrhythmias. The Holter monitoring showed less than 20 ventricular premature beats per hour and no ventricular runs. This was a group with a low risk of future arrhythmic events.

Group 3 comprised 18 healthy volunteers.

The clinical data of these patients are given in Table 1. All patients were free of antiarrhythmic drugs or $\beta$-blockers for at least five half-lives. In group 1 and 2 patients, myocardial infarction was verified by typical electrocardiographic signs (Q wave and loss of R wave) and by coronary angiography.

Statistical analyses were performed with Wilcoxon signed-rank test for unpaired data.

**Results**

**Analysis of Characteristic Patients**

Figure 4 shows the analysis of a patient after myocardial infarction with sustained ventricular
tachycardia. In the time domain, late potentials identified with a duration of 7 μV could be clearly identified after the end of the QRS complex in a high-gain recording after signal averaging (upper panel). AFD of 38 segments was calculated; for each spectrum, the computer determined the optimal order and subtracted an AFD spectrum of low order. In this patient, the spectra far outside the QRS complex (segments 1–15) were free of high-frequency (40 Hz) components. In segments at the end of the QRS complex (16–22), prominent spectral peaks in the range of 40–200 Hz were present; they disappeared in segments 27–38 of the S wave. The factor of normality was abnormal in this patient. Thus, late potentials in the time domain are reflected as high-frequency spectral peaks in segments at the end of the QRS complex but not far outside the QRS complex. AFD allowed exact determination of the late components of the QRS complex.

**Table 2. Results in Patients After Myocardial Infarction With Ventricular Tachycardia (Group 1)**

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<tr>
<th>Patient</th>
<th>Age (yrs)</th>
<th>Ejection fraction (%)</th>
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<th>Infarction site</th>
<th>Time domain</th>
<th>AFD</th>
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<td>91</td>
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<tr>
<td></td>
<td>55/F</td>
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<td>103</td>
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<td>-</td>
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<tr>
<td></td>
<td>68/M</td>
<td>48</td>
<td>115</td>
<td>A - -</td>
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<tr>
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<td>84</td>
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<td>P - y -</td>
<td>-</td>
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<td>100</td>
<td>A - y -</td>
<td>n</td>
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</table>

**Notes:** LP+, late potentials present; FFT, fast Fourier transform; NF, factor of normality; AFD, adaptive frequency determination; Loc, start and end of LP referred to at end of QRS; A, anterior infarction; P, posterior infarction; +, aneurysm present; Simson method: p, abnormal; n, normal; --, not evaluated; bold values are abnormal.
beginning and the end of the late potentials; as soon as late potentials are included in a segment, the spectral area of the corresponding spectrum increases (Figure 4, right panel). Figure 5 represents a patient after myocardial infarction who had no history of arrhythmias. He did not show late potentials in the time domain, and with AFD, the spectra were free of high-frequency components in the range above 40 Hz. The factor of normality was normal.

A difficult but realistic example in another patient after myocardial infarction with ventricular tachycardia is demonstrated in Figure 6. The electrocardiogram was of poor quality because electrode impedance was high (20 kΩ) due to obesity and dry skin, and the recording was done within the coronary care unit. In the time domain, it was impossible to distinguish between late potentials and noise. The Simson method was normal (QRS duration, 110 msec; RMS, 33 μV; Figure 6, upper right panel). Spectral analysis with AFD revealed two forms of high-frequency components—spectral peaks that were present in all segments (far outside the QRS and at the end of the QRS complex) and peaks that were present only at the end of the QRS complex but absent in segments far outside. The first were caused by noise interference that was equally distributed in all segments, and the latter were caused by delayed ventricular activation. The factor of normality was abnormal. Thus, late potentials and noise can be differentiated in the frequency domain by characteristic arrangements of spectral peaks. The Simson method was normal because the QRS limits were determined incorrectly due to the high noise level. The patient was reevaluated after discharge from the intensive care unit; then, the noise level was much lower, the high-frequency components at 180 Hz had gone, and the components at the end of QRS remained in the same frequency range. In the time domain, late potentials could now be detected, and the Simson method turned to abnormal.

Comparison of Patient Groups

Results in control group (group 3). Only two of 18 individuals had a pathological factor of normality with AFD. One of them had borderline right bundle branch block; the other presented with an incomplete right bundle branch block with a RsR' complex. In the other subjects, no spectral components in the range above 40 Hz could be seen, and only three individuals had spectral peaks in early segments (more than 20 msec inside the QRS complex). Fourier analysis was abnormal in the same two persons, and the Simson method was abnormal in three others. In the time domain, no one revealed late potentials with the visual approach; 14 leads in nine subjects could not be evaluated because of noise or steep rise of the ST segment.

Results in infarction patients without ventricular tachycardia (group 2). With AFD, the factor of normality was pathological in four of 21 patients (19%), and they had spectral peaks in segments after the end of the QRS complex. Three of them also had distinct late potentials in the time domain, and Fourier analyses were abnormal. All of them had myocardial infarction less than 1 year earlier. With the visual approach, 19 recordings could not be evaluated. The Simson method was abnormal in three of 14 patients (seven bundle branch blocks excluded); however, two of these three patients had normal results with frequency analysis. It is noteworthy that AFD demonstrated microvolt potentials within the QRS complex in 13 of 21 patients (Figure 7). They started between −30 and −20 msec and ended before the termination of the normal ventricular activation (by a mean of −6±2 msec before the end of QRS).

Results in patients with ventricular tachycardia (group 1). AFD revealed a pathological factor of normality in 28 of 38 patients (74%). There was a close relation between the leads demonstrating abnormal AFD
results and the channels with late potentials in the time domain, although a considerable number of leads could not be evaluated in the time domain (Table 2). Concordance between AFD results and analysis with Fourier transform was almost perfect. There was a discrepancy in only two patients. The Simson method was abnormal in 16 of 27 patients (11 bundle branch blocks excluded); however, in nine of 27 cases, the Simson result did not correlate with the spectral methods (Table 2). In five of nine patients, the Simson method failed because determination of the QRS limits was incorrect due to noise interference.

AFD revealed high-frequency components in a total of 61 records in 28 patients (Table 2). In eight cases, they began very early in the QRS complex (more than −20 msec with regard to the end of QRS) and ended before the termination of the QRS; the factor of normality was normal. They were not considered to be “late” potentials because such microvolt signals occurred equally frequently in patients after myocardial infarction without ventricular tachycardia. In 16 records, the spectral peaks started more than −20 msec inside the QRS but extended beyond the end of QRS. In the majority of cases, the peaks first appeared in the terminal 20 msec of the QRS and spanned into the ST segment. Only six records showed first appearance of spectral peaks after the termination of normal ventricular conduction. The mean duration of late potentials amounted to 22±10 msec after the end of the QRS complex. No correlation was found between AFD results and ejection fraction, presence of aneurysm, or site of infarction.

Figure 8 summarizes the results in our patient groups. The factor of normality was significantly lower in patients after myocardial infarction with ventricular tachycardia. There was no statistical difference between groups 2 and 3.

Analyses of segments that started more than 30 msec inside the QRS complex revealed no differences between the patient groups and might be disturbed by the very steep deflection of the R wave.

Discussion

Frequency Analysis With AFD

In principle, spectral methods offer some advantages for analysis of delayed ventricular activation compared with conventional time domain analysis; differentiation of noise is easier, and a priori filtering of the input signal is not necessary. Therefore, filter artifacts and the influence of differing cutoff points can be avoided. Fourier analysis has been the standard algorithm used for calculation of the frequency spectrum. Altered frequency components in electrocardiographic signals from patients with ventricular tachycardia could be found by some but not by others; mainly due to methodological problems. The spectrottemporal mapping with Fourier transform significantly improved the differentiation between late potentials and noise and allowed sensitive and specific determination of delayed ventricular...
However, the Fourier transform allows only a qualitative differentiation of patients (normal or abnormal) because segment sizes need to be long (80–120 msec) to achieve a reasonable frequency resolution. Furthermore, window functions have to be used to reduce spectral leakage (artificial frequencies) that may attenuate or even eliminate signals at the end of the segments and further reduce frequency resolution (effective frequency resolution of an 80-msec segment and Blackman Harris window is about 25 Hz). With fast Fourier transform, the segments must not include more than 20 msec of the QRS complex.10

In recent years, “autoregressive models” have become increasingly popular to improve frequency analysis with special emphasis on short- and slow-signal waveforms.17,23,24 They use a quite different mathematical approach to the problem; they estimate the signal components by the stepwise calculation of filter coefficients in an autoregressive loop (see “Methods”) rather than by splitting it up into sine waves like fast Fourier transform does (consequently, fast Fourier transform produces errors if a segment holds less than one period of the slowest component).16,18,22,26,27 Autoregressive models estimate and extrapolate a temporary waveform, whereas the latter requires a pseudoperiodic signal. In general, autoregressive models have a high-frequency resolution, do not require window functions, and therefore might be well suited for spectral analysis of the electrocardiogram.

There are different autoregressive models. The Burg algorithm, also known as the maximum entropy method (MEM), was originally designed for analysis of very slow waveforms in geophysics6,8,28–31 but has also been used in techniques7 and biomedicine.9 In our experience, MEM is a very sensitive method for the analysis of the electrocardiogram with high-frequency resolution, but the determination of the optimal order is a critical problem,22,32–35 and the robustness of the algorithm is somewhat limited by line-splitting (one spectral peak may be split into two or three components).36–39 Artificial peaks, and phase bias,17,30 AFD has not been applied to electrocardiographic signals until now; it uses a similar approach with different mathematical equations to calculate the filter coefficients and with the fundamental difference that the order of the model is calculated before autoregression is started (MEM increases the order until a stop criterion is met).9,15,19,40,41 Spectral analysis of the same signals with MEM and AFD (test signals and electrocardiogram; not shown) revealed that AFD results are more stable and that line-splitting does not occur. Artificial peaks (a critical weakness of autoregressive models like MEM) were rare and of low amplitude (<20 dB compared with peaks due to late potentials) and did not disturb the analysis of the signals of interest; this was due to the characteristics of the algorithm itself and the strict criteria to calculate the optimal order. As a consequence, false abnormal results were rare in patient groups 2 and 3 without arrhythmias. If the late potentials were eliminated in the time domain with the computer cursor, the high-frequency peaks of the AFD spectra disappeared in all patients in whom this was applicable. The main advantages of Fourier transform are very high-frequency resolution in segments as short as 25 msec (with electrocardiographic data, approximately 6 Hz; fast Fourier transform, >40 Hz at the same sample size), that there is no need of window functions, that spectral leakage is insignificant, and that the influence of underlying slow waves as the ST segment is small. The autocorrelative mode of operation of the algorithm makes AFD less susceptible to noise; spectrottemporal mapping with AFD significantly improves the discrimination of noise, which can be identified by spectral peaks present in all segments (Figure 6). The calculation of “difference spectra”—not possible with fast Fourier transform—reduces influences of slow components such as the fundamental ST segment and thereby enhances the graphical display and the “postprocessing” of the data (spectral area for localization and factor of normality) and is the prerequisite for analysis of high-frequency components within the terminal QRS. The calculation of difference spectra might create errors,42 but for the analysis of delayed ventricular activation, these proved to be insignificant because of the great difference in amplitude and frequency of the fundamental ST wave and the late potentials.

**Clinical Relevance of AFD**

With respect to the capability to identify late potentials and to differentiate noise interference, AFD of 25 msec segments yielded results equal to those of Fourier transform analysis of 80-msec segments. The shorter segment size is beneficial because the segments center on the point of interest and are not distorted by unwanted signal components. AFD allowed a much more accurate evaluation of late potentials; their frequency content could be checked more precisely because the corresponding spectral peaks were narrow and did not merge (better frequency resolution compared with fast Fourier transform). Unlike fast Fourier transform, the beginning and end of the late potentials could be precisely determined, information that is otherwise lost in the frequency domain. The terminal portion of the QRS may also be analyzed; this is not possible in the time domain without the limitations of complex band-pass filtering or with fast Fourier transform because steep slopes of the QRS cause spectral artifacts. Our results indicate that microvolt signals occur not only in infarction patients with VT but also in a considerable number of patients after myocardial infarction without ventricular tachycardia and, rarely, in healthy persons. In infarction patients without arrhythmias, the high-frequency components occurred more often inside the QRS (Figure 7), and the pathological feature seems to be the amount of delay of these deflections rather than the fact of their
existence. However, a clear statement about these components is impossible because this phenomenon cannot be studied with previous methods at the body surface. A methodological artifact is unlikely because those components do not occur in test signals and only rarely in healthy subjects. Follow-up of these patients will show whether high-frequency components within the QRS have a prognostic significance with regard to future arrhythmic events.

AFD combines the advantages of frequency domain analysis (good differentiation between noise and late potentials and high-pass filtering not necessary) and time domain analysis (exact localization of late potentials). In a significant number of patients, time domain analysis was not possible or was incorrect (Figure 6), but AFD allowed correct classification of the patient. Like in Fourier transform, patients with bundle branch block need not be excluded with AFD; therefore, the clinical applicability and sensitivity is significantly improved. The sensitivity of the Simson method was only 42% in patient group 1, mainly because subjects with bundle branch block (a substantial subgroup of patients after myocardial infarction) in general had to be excluded. Unlike the Simson method, the definition of abnormality does not depend on the noise level. With AFD, a single parameter, the factor of normality, is used to classify the patients. With Fourier transform, this factor was determined by cross correlation of the spectra; with AFD, the spectral area was used because a small shift of the narrow peaks caused the correlation to decrease to zero.

Limitations and Conclusion

AFD is a powerful algorithm for precise evaluation of late potentials in the frequency domain. Unlike fast Fourier transform, spectral peaks are narrow and do not merge into one another because of poor frequency resolution. The purpose of this retrospective study was to validate the method and determine characteristics in each patient group. With the method of AFD, the incidence of late potentials should be prospectively studied in larger groups of patients to assess its prognostic value. The method also offers promise in the areas of influence of antiarrhythmic drug therapy on late potentials, changes of delayed ventricular activation during programmed ventricular stimulation, and late potentials in disorders other than coronary artery disease.

Does AFD take the place of fast Fourier transform? We do not think so; fast Fourier transform is a quick and reliable method for determining whether late potentials are present. AFD is useful for subsequent detailed analysis; however, the clinical value should be tested in a larger study population.

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Appendix

To fully understand and reproduce the method, the following references are recommended. For information on the principles of autoregressive spectral analysis, refer to Childers et al (pp 34–41 and 42–45), Kay and Marple (pp 1380–1419), and Marple (pp 441–454). For information on the AFD algorithm, refer to Kalouptsidis (pp 661–670) and Simson (pp 699–709). And for information on determination of the order, refer to Akaike (pp 203–217), Akaike (pp 716–723), Burg (pp 26–32), and Broersen (pp 874–879).

References

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Top-resolution frequency analysis of electrocardiogram with adaptive frequency
determination. Identification of late potentials in patients with coronary artery disease.
R Haberl, H F Schels, P Steinbigler, G Jilge and G Steinbeck

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