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Baseline reconstruction for localization of rapid ventricular tachycardia from body surface potential maps

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Abstract

Determination of an accurate electrocardiographic (ECG) baseline is generally needed for localization of ventricular arrhythmias with body surface potential mapping (BSPM). We suggest a novel signal processing method for ECG baseline reconstruction during monomorphic ventricular tachycardias (VT). The method is based on an assumption that VT consists of similar ventricular extrasystolic beats with overlapping depolarization and repolarization. The sequential reconstruction algorithm utilizes information of small variations in the heart rate and yields a non-overlapping QRST-signal, provided that the measurement set-up has a high enough temporal resolution to avoid distortions due to sampling differences and misalignment of individual beats. The reconstructed QRST-signal is utilized to subtract overlapping T-waves from the QRS complexes during VT. The use of the method is demonstrated with clinically measured BSPM data.

Keywords: electrocardiography, baseline, ventricular tachycardia, body surface potential mapping

1. Introduction

Catheter ablation of ventricular tachyarrhythmias requires intracardiac mapping and pacing in an electrophysiological study (EPS) to locate the arrhythmogenic tissue. Standard 12-lead electrocardiograms (ECGs) are used to guide the catheter positioning (Josephson *et al* 1981).

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Non-invasive localization accuracy of arrhythmia sources can be improved by body surface potential mapping (BSPM) (SippensGroenewegen *et al* 1993). The BSPM localization of arrhythmia sources is usually based on previously recorded databases of ventricular arrhythmias (e.g., SippensGroenewegen *et al* 1990, 1992, Green *et al* 1994, Molin *et al* 1997). In BSPM database localization, the ECGs obtained during arrhythmia are averaged at each channel over the time interval from Q-wave onset to the S-wave offset (QRS integral) and the result is plotted as a distribution on the body surface (SippensGroenewegen *et al* 1990, Simelius *et al* 1996a).

In rapid ventricular tachycardia (VT), the electrocardiographic (ECG) T-wave and the next QRS-complex overlap, and no clear isoelectrical baseline is visible. In conventional ECG methods for the source localization of VT (Josephson *et al* 1981, Waxman and Josephson 1982), the exact baseline is not needed for morphological analysis of QRS-complexes. On the other hand, when applying computational methods such as localization of arrhythmogenic source by databases or by solution of the ECG inverse problem (e.g., Horáček and Clements 1997, van Oosterom 1999, Nenonen *et al* 2001, Tilg *et al* 2002, 2003), an accurate baseline of ECG is needed because the analysis is based on the absolute potential values generated by the cardiac activation. The large number of channels in BSPM (32–256) motivates the use of computerized analysis methods. Correctly analysed, the BSPM acquired during an EPS study can be used to guide the catheter ablation (SippensGroenewegen *et al* 1993, Potse *et al* 2000). Limitations for the routine use of BSPM in a catheterization laboratory are the lack of standardization and still ongoing development of new analysis methods.

The ECG baseline is traditionally determined visually. In earlier studies (Linnenbank 1991, Kemmelings *et al* 1994) the exact determination of electrocardiographic baseline was not found crucial in BSPM. However, we feel that the solution of inverse problems or comparison of isopotential maps from different beats requires reliable determination of the baseline to achieve reproducible and consistent results. Typical ECG baseline correction methods utilize filtering (Pottala *et al* 1989, Sörnmo 1993) or spline interpolation methods (Froning *et al* 1988) to reduce baseline wandering. Such methods are efficient when the baseline disturbances arise from external sources, such as patient movement, electrode potential drift or ambient electromagnetic fields. An isoelectrical time interval in the cardiac cycle has to be found first (i.e. a period with no cardiac electrical activation). However, the heart rate in life-threatening VTs increases drastically and no isoelectrical segments can be found in the ECGs. Furthermore, attempts to use high-pass filtering in such VTs will shift the zero-level to the average of the QRS amplitude in each lead.

We have proposed a recursive baseline reconstruction method for VT, by extracting information of repolarization from the RR-variations in the ECG data (Jokiniemi *et al* 1997). In the present work we describe an improved non-recursive method, which is computationally efficient and robust against timing errors and measurement noise. We also report the performance of the new method in BSPM recordings of clinical VT patients. The goal for ECG baseline reconstruction of VT is to make the BSPM analysis more reliable and reproducible in clinical settings.

2. Methods and materials

2.1. Basic assumptions

The depolarization in VT starts typically from a border zone between infarcted and normal myocardial tissue. The depolarization wavefront does not initiate along conduction pathways but propagates directly through the myocardium from an abnormal origin site. Due to a short

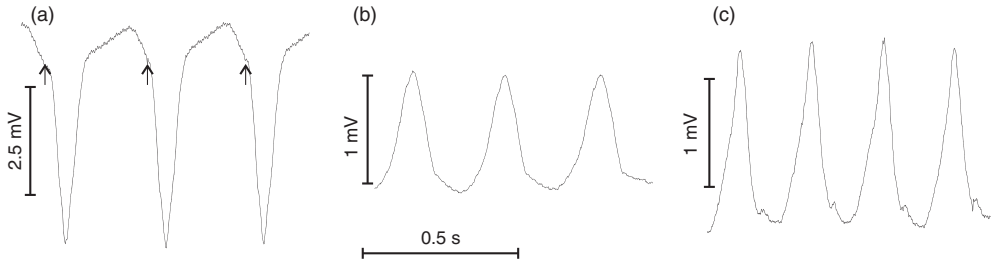

Figure 1. Ventricular tachycardia beats in BSPM (Simelius *et al* 1996b) recorded in the catheterization laboratory of the Helsinki University Central Hospital. There are no isoelectrical moments since the onset of QRS is distorted by overlapping T-wave. (a) Ventricular pacing from channel 46. The pacing instants are denoted by the arrows. (b) Programmed VT with the rate of 210 bpm, channel 81. (c) Ventricular tachycardia with the rate of 270 bpm, channel 55.

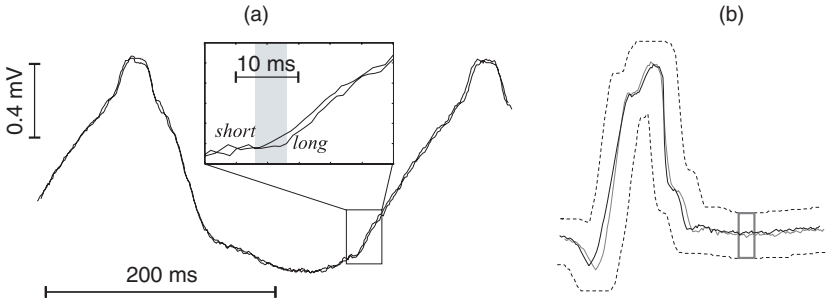

Figure 2. (a) Two separate time intervals of VT (lead aVL) triggered from the maximal slope of the first beat. The close-up shows the additional T-wave information in the shaded RR-variation zone. (b) The selection of beats in signal averaging is based on whether they fit inside a predefined 'tube'. It is an envelope formed by sliding a box over the cumulative average (Paavola *et al* 1995). Here the width and height of the box are, respectively, 10 ms and 0.1 mV.

cycle length of VT, a new depolarization wavefront (QRS) starts before the myocardium has completely repolarized. Thus in body surface recordings, such as 12-lead ECG, BSPM and magnetocardiography (MCG), the observed signal is a superposition of the fields generated by both depolarization and repolarization. In such a case, a direct determination of ECG baseline from body surface recordings becomes impossible.

The overlapping of initial QRS-complex with the terminal part of the preceding T-wave (figure 1) is thus not caused by overlapping phenomena at the cellular level, but by spatial summation of the electromagnetic fields. In principle, it is possible to separate these two activation components in surface recordings, and to obtain an ECG baseline from the time interval before the onset of QRS without distortion caused by preceding T-wave. Such baseline extraction is also of clinical importance since most BSPM localization approaches rely on the baseline. Furthermore, databases for comparison of rapid VTs have been gathered using slow pacing, where the preceding T-wave causes no distortion of initial QRS (Waxman and Josephson 1982, SippensGroenewegen *et al* 1990, 1992).

2.2. RR variations

When examining two overlaid beats of VT triggered from the maximum slope of previous QRS-complex (figure 2(a)), it is notable that the beat with a long preceding RR-interval

exhibits more pure T-wave before the onset of the QRS-complex than a short RR-interval beat. By RR-interval we mean here the trigger-to-trigger distance when triggering is performed by detecting the leading edge of QRS-complex. As a result, the length of the preceding RR-interval can be used as a measure of the amount of pure non-overlapping T-wave. When taking two beats with maximal difference in the preceding RR-interval, the length of the pure T-wave, not contaminated by initial QRS-complex, is also maximal in the RR-variation zone (shaded in figure 2(a)).

We assume that the continuous VT signal $S(t)$ is a composition of similar overlapping VESs $V(t)$ and noise $E(t)$

$$S(t) = \sum_{i=1}^{\infty} V\left(t - \sum_{j=1}^{i} RR_j\right) + E(t) \tag{1}$$

where RR_j is the RR-interval between the beats $j-1$ and j. Taking two windows of VT, each window containing signal only from two VESs with different RR-intervals, we have

$$\begin{aligned} S_s(t) &= V(t) + V(t - RR_s) + E(t) \\ S_l(t) &= V(t) + V(t - RR_l) + E(t) \end{aligned} \tag{2}$$

where indices s and l refer to the shorter and longer RR-intervals in these signals. Note that equations (1) and (2) are not limited to ECG signals only. In principle, they can be applied in other studies dealing with the superposition of pulses, such as recordings of neural action potential trains. In the following we, however, focus on an ECG reconstruction method based on the assumption of equation (2).

2.3. Baseline reconstruction method

We have previously developed a recursive reconstruction method based on the basic assumption stated in equation (1) (Jokiniemi *et al* 1997). Next we propose a non-recursive derivative-based approach which is faster and more straightforward to implement in a practical set-up than the recursive approach.

Subtracting two beats as defined by equation (2) with different RR-intervals from each other yields

$$S_s(t) - S_l(t) = V(t - RR_s) - V(t - RR_l) + \sqrt{2}E(t). \tag{3}$$

By the definition of a discrete derivative by difference quotient it is clear that the subtraction is actually a discrete N-point derivative of $V(t)$

$$S_s(t) - S_l(t) = N \cdot \frac{\mathrm{d}}{\mathrm{d}t} V(t - RR_d) + \sqrt{2}E(t) \tag{4}$$

where $N = RR_l - RR_s$ and $RR_d = \frac{RR_s + RR_l}{2}$. Integrating the previous equation yields

$$\begin{aligned} \int S_s(t) - S_l(t)\,\mathrm{d}t &= \int \left[N \frac{\mathrm{d}}{\mathrm{d}t} V(t - RR_d) + \sqrt{2}E(t) \right] \mathrm{d}t \\ &= NV(t - RR_d) + \sqrt{2} \int E(t)\,\mathrm{d}t. \end{aligned} \tag{5}$$

Using the previous equations, the determination of the estimate for $V(t)$ is a straightforward task

$$\hat{V}(t - RR_d) = \frac{1}{N} \int [S_s(t) - S_l(t)]\,\mathrm{d}t. \tag{6}$$

In this approach the noise does not accumulate, but is scaled by a factor $\sqrt{2}/N$ and integrated over time. The nature of sinusoidal interference such as 50/60 Hz line noise does not change in this procedure, but the white noise transforms into $1/f$ noise due to integration in equation (5).

In this method the N is basically not limited to an integer. The reconstruction by integration is not limited by a step of integer size as in the recursive method (Jokiniemi *et al* 1997). The most straightforward way of determining the scaling factor $1/N$ relies on the determination of RR_s and RR_l. We have also applied more sophisticated scaling methods, which allow real values of N. Triggered beats are temporally aligned by sliding two beats with respect to each other prior to computing the estimate $\hat{V}(t)$. The correct estimate can be selected by minimizing the signal power in the preceding T-wave area. The value of N can be estimated by minimizing the signal power in the QRS area, where no previous T-wave is present. This estimation also results in a correct estimate for the preceding T-wave. In addition, weighted signal interpolation can be used in these minimizations to improve the results.

The difference between RR_s and RR_l does not have any effect on the algorithm parameters as in the recursive method and we can always select $n \cdot (n-1)/2$ beat pairs, where n is the number of beats in continuous VT with a known RR-value. From $n \cdot (n-1)/2$ estimates we can simply select the complexes with the highest signal-to-noise ratio requiring all data points to be inside a predefined range of variation on the selected interval around the median value of individual beats as illustrated in figure 2(b) (Paavola *et al* 1995, Väänänen *et al* 2000). The selected beats are then averages to reduce noise and possible reconstruction errors. Due to the number of estimates being $O(n^2)$, the signal-to-noise ratio is enhanced by a factor proportional to n compared to $\sqrt{n}$ of conventional averaging.

2.4. Verification of the reconstruction results

The reconstructed VES:s are verified by regenerating the VT using the previously computed estimates. The VES:s are overlapped at correct RR-intervals and the resulting estimated VT is compared with the true signal. A residual signal is constructed by subtracting the estimated VT from the true VT and the power of this signal can be used as a measure of goodness for reconstruction. Also, visual inspection of this residual clearly reveals possible reconstruction errors.

A more sophisticated way of measuring goodness of reconstruction would be to study the spectrum of the residual signal. Normal white noise, $1/f$ noise and line interference are clearly separable from frequencies in the range of heart rate (HR) during VT. These HR-related frequencies can be described as reconstruction errors, whereas other signal components are related to external noise or interference.

2.5. Test data

We have previously used both simulated and paced BSPM data to test and demonstrate the recursive reconstruction algorithm. The same computer-generated VES used in Jokiniemi *et al* (1997) was also employed to test the new baseline reconstruction method.

Ventricular pacing can be used to produce a BSPM data resembling VT with controlled timing and RR-interval variation. In this study, pacing with varying coupling interval 300–340 ms was performed during standard EPS. A signal from one BSPM lead recorded during pacing is shown in figure 1(a). Even though the rate is notably slower than usually in VT, there is no silent isoelectrical moment in the ECG.

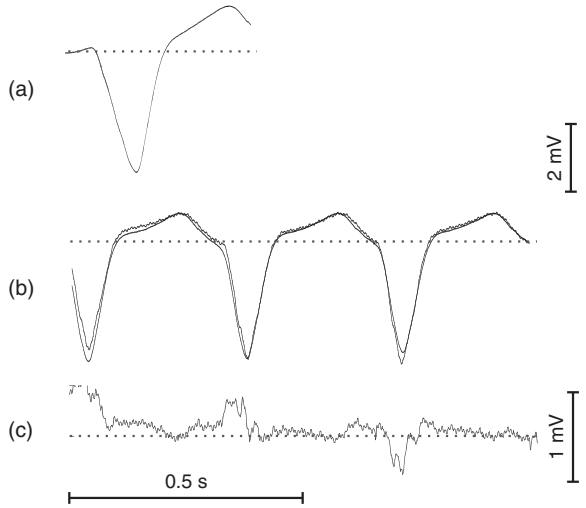

Figure 3. Reconstruction results of ventricular pacing. (a) The VES reconstructed from ventricular pacing. (b) Regenerated and original continuous signals overlapped. (c) Signal residual between regenerated and original signals. The RMS intensity of the residual signal is 140 μV.

In addition, we employed BSPM data recorded during sustained VT that was induced by programmed stimulation (Wellens 1978) from the right ventricular outflow tract. The heart rate was 210 bpm and the patient was haemodynamically stable during the VT. A part of the recorded signal is presented in figure 1(b).

As a worst-case example another VT recorded by BSPM during standard EP-study was used. A rapid VT (270 bpm) was induced by extrastimulus pacing. After 45 s the VT deteriorated into ventricular fibrillation, which was defibrillated back to sinus rhythm. One BSPM lead of the measured signal is depicted in figure 1(c). The low amplitude spikes after the QRS-complexes are due to attempted pacing for turning the VT into sinus rhythm. The pacing was unsuccessful and has no effect on the ECG except for these spikes. This ECG example demonstrates that the determination of the electrocardiographic baseline by traditional methods is impossible.

The final test for the clinical applicability of these methods is to determine how they affect the isointegral or isopotential maps. Especially when comparing two isointegral maps with each other, the baselines of these two maps have to be accurately determined.

3. Results

The reconstruction method could determine the baseline from the simulated VT data almost perfectly. The results are not shown here because they are close to the results of our earlier study (Jokiniemi *et al* 1997). It was, however, notable that the non-recursive method is more robust against triggering inaccuracies and white noise.

The reconstructed VES of the paced VT (figure 1(a)) and the regenerated pacing signal with residual are presented in figure 3. The reconstruction result of the VES is smooth and has no reconstruction artefacts. The reconstructed pacing is partly misaligned, due to which the residual has some signal components left. The derivative method performed better than the recursive method; the corresponding residual signal powers were 140 μV and 180 μV.

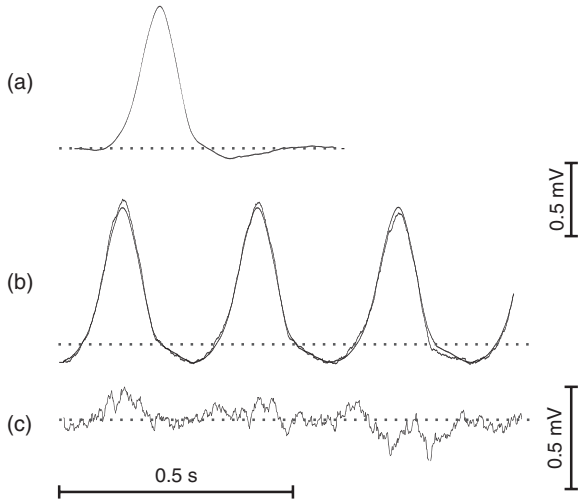

Figure 4. Reconstruction results of VT. (a) The VES reconstructed from ventricular tachycardia. (b) Regenerated and original continuous signals overlapped. (c) Signal residual between regenerated and original signals.

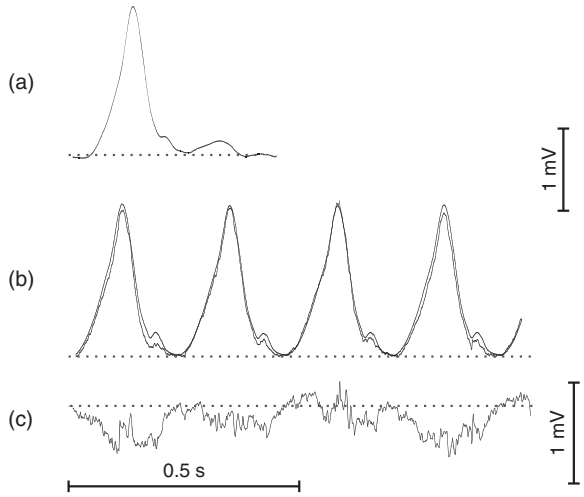

Figure 5. Reconstruction results of rapid VT (270 bpm). (a) The VES reconstructed from ventricular tachycardia. (b) Regenerated and original continuous signals overlapped. (c) Signal residual between regenerated and original signals.

The reconstructed VES of the VT (figure 1(b)) as well as the regenerated and residual signals are presented in figure 4. The reconstructed VES is free of artefacts and the baseline as well as the onset and offset of the QRS-complex are easily determinable. The regenerated VT has an identical waveform to the original signal and the residual signal has no significant signal components left. The residual signal shows how white noise transforms into $1/f$ noise in integration. The RMS amplitude of the residual is only about half of the residual RMS amplitude in the recursive method (21 μV versus 39 μV).

The reconstruction results of the faster VT (figure 1(c), 270 bpm) are depicted in figure 5. The regenerated VT shows a similar waveform compared to the original signal, even the spikes

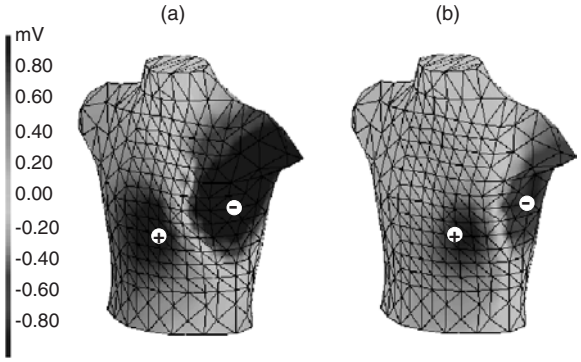

Figure 6. The resulting isointegral maps of VT, (a) after baseline reconstruction and (b) when the baseline was selected by visual inspection.

of the attempted pacing are regenerated from the reconstructed VESs. The low amplitude structure before the QRS-complex of the VES is due to these pacing spikes. The residual signal again shows characteristic $1/f$ behaviour. The RMS amplitude of the residual signal is notably lower than in the recursive method (72 μV versus 108 μV).

Finally, to visualize the baseline effect on BSPM patterns we used the data of the sustained VT case. Integral values of the QRS-complex were first computed from all 123 channels when the baseline level was set by visual determination of the QRS onset and offset. For comparison, we evaluated the QRS integral values also after the derivative-based reconstruction using the resulting QRS onset and offset (figure 4(a)). The spatial distributions of the QRS integrals are depicted in figure 6.

4. Discussion

4.1. Need of baseline

The use of BSPM in the catheterization laboratory during EP-studies aims at shortening the invasive study and at improving the success rate of RF-ablations. It has been shown that BSPM can be used to localize the origin of VT and to guide the search for the arrhythmogenic source by pace mapping more efficiently than standard 12-lead ECG (SippensGroenewegen *et al* 1993, Simelius *et al* 1996a).

BSPM localization of the source of VT is based on the interpretation of potential distribution on the body surface caused by the myocardial depolarization. This is why an accurate electrocardiographic baseline is needed. As was shown above (figure 6), the visual selection of a baseline may introduce distortions in the isointegral maps. The map creation also becomes more reproducible and accurate with automatic baseline reconstruction methods than without. The proposed baseline reconstruction method is intended to improve the accuracy of initial localization of the arrhythmogenic source in VT. The method can also make the localization procedure faster and more automatic.

We assumed each beat of VT to be similar, but the timing and overlapping of the QRS-complex and the T-wave vary. This assumption limits the application of our method only to monomorphic VTs. On the other hand, the goal is to locate the sources of arrhythmogenic activity (Potse *et al* 2000, Simelius *et al* 1996a) for the ablation procedures while polymorphic VTs may not have a single discrete source. Thus, there may not be such a need to determine the ECG baseline of polymorphic VTs.

4.2. Reconstruction

Monomorphic VT has only small RR-variation, usually around 10 ms, which makes the step in the recursive reconstruction algorithm quite short. As seen in reconstruction results of both simulated and true signals, even small timing errors can cause significant artefacts and distortions in the resulting VES (Jokiniemi *et al* 1997). The derivative method is less sensitive to noise accumulation, smaller amount of RR-variation and triggering accuracy. This method is also computationally more straightforward to implement. The results of this study show that this method is more accurate and practically more feasible. This is also supported by the computing time of 56 s (derivative) versus 43 s (recursive) with 5 s interval of 123-channel BSPM recording of VT, even if the derivative method uses 15 beat pairs compared to 4 pairs used by the recursive method (MATLAB code, 200 MHz Pentium).

Examining equation (4) it becomes clear that low-frequency baseline drift causes unwanted DC-components in the discrete derivative of reconstructed VES. This in turn generates severe linear drift in resulting VES, which cannot be easily removed. To assure errorless reconstruction, we should either extend the recording bandwidth nearly to DC or correct the baseline to some predetermined value. In the estimation of the baseline for correction we have successfully used cubic spline interpolation (Froning *et al* 1988) between T-wave maxima or minima. The peak of the T-wave was selected as a pseudo-baseline time reference, because at that moment there is electrical activation of only one VES.

Due to integration in equation (6) a sinusoidal signal, such as 50/60 Hz line interference, will be scaled by its angular frequency. The effect of line interference could be minimized by decimating the input signal before reconstruction to the lowest acceptable sampling frequency. When decimating the signal the temporal resolution decreases, which might be accepted if high enough noise reduction is achieved.

4.3. Hardware limitations

The equipment used in recording VT has a notable effect on the timing accuracy. As seen in the present and previous studies (Jokiniemi *et al* 1997), the temporal resolution required for errorless reconstruction is of the order of 1 ms, which is possible only if the sampling frequency in the measurement hardware is at least 1 kHz. Excessive lowpass filtering must also be avoided to prevent signal distortion, which would make accurate triggering even more difficult. Temporal interpolation can make the accuracy slightly better, but one must remember that it does not create any new information. Even if the derivative method is more insensitive to triggering, the correct derivative from equation (4) can only be computed if the signals have no temporal distortion in them.

5. Conclusions

The new reconstruction method enables determination of the baseline and the onset and offset of QRS-complexes of rapid VT. Temporal resolution and precision of the beat alignment are the main restrictions in the accuracy and the applicability. Even if the reconstruction results of ventricular pacing and VT are not perfect, the method improves significantly the baseline detection.

In addition, we demonstrated that the method can improve the clinical applicability of the body surface potential mapping in localization of rapid ventricular tachycardia on the basis of reliable isopotential map creation. The reproducibility of accurate isopotential maps, e.g.,

for database matching, would be limited without sophisticated signal processing such as the method applied in the present study.

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References

Froning J N, Olson M D and Froelicher V F 1988 Problems and limitations of ECG baseline estimation and removal using a cubic spline technique during exercise ECG testing: recommendations for proper implementation *J. Electrocardiol.* **21** 149–57

Green L S, Lux R L, Ershler P R, Freedman R A, Marcus F I and Gear K 1994 Resolution of pace mapping stimulus site separation using body surface potentials *Circulation* **90** 462–8

Horáček B M and Clements J C 1997 The inverse problem of electrocardiography: a solution in terms of single- and double-layer sources on the epicardial surface *Math. Biosci.* **144** 119–54

Jokiniemi T, Simelius K, Nenonen J, Toivonen L and Katila T 1997 An electrocardiographic baseline reconstruction method for rapid ventricular tachycardia *Computers in Cardiology* (Piscataway, NJ: IEEE) pp 251–4

Josephson M, Horowitz L, Waxman H, Cain M, Spielman S and Greenspan A 1981 Sustained ventricular tachycardia: role of 12-lead electrocardiogram in localizing site of origin *Circulation* **64** 257–72

Kemmelings J G, Linnenbank A C, Muilwijk S L, SippensGroenewegen A, Peper A and Grimbergen C A 1994 Automatic QRS onset and offset detection for body surface QRS integral mapping of ventricular tachycardia *IEEE Trans. Biomed. Eng.* **41** 830–6

Linnenbank A C 1991 Effect of baseline correction on QRS integral patterns in body surface maps of ventricular tachycardia *Proc. 13th Int. Conf. IEEE/EMBS* pp 789–90

Molin F, Savard P, Dubuc M, Kus T, Tremblay G and Nadeau R 1997 Spatial resolution and role of pacemapping during ablation of accessory pathways *PACE* **20** 683–94

Nenonen J, Pesola K, Hänninen H, Lauerma K, Takala P, Mäkelä T, Mäkijärvi M, Knuuti J, Toivonen L and Katila T 2001 Current-density estimation of exercise-induced ischemia in patients with multivessel coronary artery disease *J. Electrocardiol.* **34** (*suppl.*) 37–42

Paavola M, Simelius K, Montonen J, Nenonen J and Katila T 1995 Averaging in cardiographic signal processing *Proc. Finnish Signal Processing Symp. (FINSIG'95)* 138–41

Potse M, Linnenbank A C, Peeters H A, SippensGroenewegen A and Grimbergen C A 2000 Continuous localization of cardiac activation sites using a database of multichannel ECG recordings *IEEE Trans. Biomed. Eng.* **47** 682–9

Pottala E W, Bailey J J, Horton M R and Gradwohl J R 1989 Suppression of baseline wander in the ECG using a bilinearly transformed, null-phase filter *J. Electrocardiol.* **22** 243–7

Savitzky A and Golay M 1964 Smoothing and differentiation of data by simplified least squares procedures *Anal. Chem.* **36** 1627–39

Simelius K, Jokiniemi T, Nenonen J, Tierala I, Toivonen L and Katila T 1996a Arrhythmia localization using body surface potential mapping during catheterization *Proc. 18th Int. Conf. IEEE/EMBS* pp 975–6

Simelius K, Tierala I, Jokiniemi T, Nenonen J, Toivonen L and Katila T 1996b A body surface potential mapping system in clinical use *Med. Biol. Eng. Comput.* **34** 107–8

SippensGroenewegen A, Hauer R N W, van Hemel N M, Kingma J H, de Bakker J N, Janse M J and Robles de Medina E O 1993 Design and clinical application of a body surface mapping reference data base for detailed localization of ventricular tachycardia foci in patients without structural cardiac disease *Cardiac Mapping* ed M Shenasa, M Borggrefe and G Breithardt (Mount Kisco, NY: Futura) pp 347–63

SippensGroenewegen A, Spekhorst H, van Hemel N M, Kingma J H, Hauer R N W, Janse M J and Dunning A J 1990 Body surface mapping of ectopic left ventricular activation: QRS spectrum in patients without structural heart disease *Circulation* **82** 879–96

SippensGroenewegen A, Spekhorst H, van Hemel N M, Kingma J H, Hauer R N W, Janse M J and Dunning A J 1992 Body surface mapping of ectopic left ventricular activation: QRS spectrum in patients with prior myocardial infarction *Circ. Res.* **71** 1361–78

Sörnmo L 1993 Time varying digital filtering of ECG baseline wander *Med. Biol. Eng. Comput.* **31** 503–8

Tilg B, Fischer G, Modre R, Hanser F, Messnarz B, Schocke M, Kemser C, Berger T, Hintringer F and Roithinger F 2002 Model based imaging of cardiac electrical excitation in humans *IEEE Trans. Med. Imag.* **21** 1031–9

Tilg B, Fischer G, Modre R, Hanser F, Messnarz B and Roithinger F 2003 Electrocardiographic imaging of atrial and ventricular electrical activation *MEDIA* at press

Väänänen H, Korhonen P, Montonen J, Mäkijärvi M, Nenonen J, Oikarinen L, Simelius K, Toivonen L and Katila T 2000 Non-invasive arrhythmia risk evaluation in clinical environment *Herzschrittmacher Elektrophysiol.* **11** 229–34

van Oosterom A 1999 The use of the spatial covariance in computing pericardial potentials *IEEE Trans. Biomed. Eng.* **46** 778–87

Waxman H L and Josephson M E 1982 Ventricular activation during ventricular endocardial pacing I: Electrocardiographic patterns related to the site of pacing *Am. J. Cardiol.* **50** 1–10

Wellens H J 1978 Value and limitations of programmed electrical stimulation of the heart in the study and treatment of tachycardias *Circulation* **5** 845–53