Reproducibility of the signal-averaged electrocardiogram using individual lead analysis

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ABSTRACT

The aim of the study was to evaluate the immediate reproducibility of time domain parameters in the signal averaged electrocardiogram using a new method for endpoint determination in individual Frank leads. The method is based on a statistical model of the electrocardiogram (ECG) to which maximum likelihood (ML) estimation is employed. The reproducibility of the ML method was compared to that of conventional time domain analysis using the vector magnitude (VM) of Frank leads. Fifty-nine patients were included in the study and two consecutive ECGs were recorded for signal averaging. The results showed that the mean of the absolute difference of the filtered QRS duration (QRSD) between two consecutive recordings was significantly lower for the ML method than the conventional method when employing 60 Hz highpass filtering (2.1±2.2ms vs 5.9±10.2ms, $p<0.05$). Moreover, the ML method resulted in significantly longer QRSD compared to the VM-based method ($p<0.05$). The terminal amplitude of the QRS complex (RMS40) showed a greater variability than for QRSD for both methods although the ML method was associated with a higher reproducibility than for the VM method for the 60 Hz filter. These findings may contribute to a better identification of patients at high risk for ventricular arrhythmias. Moreover, a reduction of the measurement errors has important implications when changes of the QRS is analysed over time.

INTRODUCTION

Ventricular late potentials which are thought to originate from areas with nonhomogenous conduction properties occur as low voltage and high frequency components at the end of or after the normal QRS complex.[1] Signal-averaged electrocardiography is a non-invasive technique used to detect these late potentials from the body surface. Although clinical utility of the technique has been subject to thorough studies during the last decade,[2-6] it has mainly been used to predict the spontaneous occurrence of ventricular tachyarrhythmias or sudden death in patients after myocardial infarction.[7,8] Moreover, several studies have tried to correlate dynamic changes in the signal-averaged electrocardiographic (SAECG) parameters to the effect of antiarrhythmic surgery[9,10] and antiarrhythmic drug efficacy[11] and to assess the effect of thrombolytic therapy on evolution of the arrhythmogenic substrate.[12,13] It is thus of great clinical importance to determine the reproducibility of the SAECG for an adequate evaluation of dynamic changes in the measurements. Although some investigators have reported a high reproducibility of the time domain analysis of SAECG,[14-16] others have found that the technique lacks sufficient reproducibility of the results.[9,17,18]

The methods employed for analyzing SAECG do not allow an optimal detection of localized delayed conduction. Rather large discrepancies between the latest electrical activity recorded from the body surface and that recorded directly from the epi- and endocardium has been reported.[19-21] Thus, late potentials of short duration or those originating from an early depolarized area of the heart can be missed using the body surface ECG, which can be related to unacceptably high noise levels as well as to the computation of the vector magnitude (VM) per se. It is therefore important to develop a method which can improve the sensitivity without loss of specificity, especially in difficult cases such as anterior myocardial infarction.

The aim of the present study was to evaluate the immediate reproducibility of individual lead analysis using a new method for endpoint determination. The reproducibility of the new method was compared to that obtained by conventional time domain analysis using the VM. The effect of different filter cut-off frequencies was also investigated.

MATERIAL AND METHODS

Patient population

Fifty-nine patients (7 women and 52 men) with a mean age of 58 ± 16 years (range 23 to 79 years) who underwent recording of SAECG were included. There were 44 patients with the previous myocardial infarction (20 anterior, 14 inferior, 4 both and 6 non specified location), 7 with arrhythmogenic right ventricular dysplasia and 8 with non-ischemic ventricular arrhythmias. Thirty-four patients had suffered from sustained ventricular tachycardia and 3 patients from nonsustained ventricular arrhythmias. Exclusion criteria included: complete bundle branch block, atrial fibrillation and ventricular preexcitation on the resting ECG, a noise level in the SAECG exceeding 0.8 μV or a myocardial infarction within 1 week of data acquisition.

Signal averaging

Data acquisition. The ECG signal was acquired during eight minutes in an unshielded room using standard Frank X, Y and Z leads[22]. Nine electrodes were used to record eight leads from which the three orthogonal leads X, Y and Z could be calculated in full accordance with the equations given by Frank[23]. The five chest electrodes, denoted A, C, E, I, M, were positioned at the same horizontal level (fourth intercostal space). The electrodes A and I were positioned in the left and right midaxillary lines, respectively. The electrodes E, M and C were positioned on the sternum, spine and in the left-anterior oblique position between the electrodes A and E, respectively. The remaining four electrodes were positioned on the left arm (LA), right arm (RA), left leg (LL) and on the back of the neck (H). The central terminal (CT) was calculated by (LL+LA+RA)/3. The eight input leads were (I-CT, E-CT, C-CT, A-CT, M-CT, H-CT, LA-RA, LL-RA). The signal was digitized at a sampling rate of 1000 Hz with an amplitude resolution of 0.6 μV (equipment by Siemens-Elema AB, Solna, Sweden).

Analysis. The ECG recording was analyzed in an off-line mode using software developed by the authors. The analysis included beat alignment in which each bandpass filtered beat was cross-correlated to a template beat (filter cut-off frequencies at 5 and 35 Hz). The beat was shifted until the highest cross-correlation value was found. Beats with a correlation below 0.985 were excluded from averaging. The number of beats accepted for averaging was typically in the range from 400 to 500 beats. The data was then split into two parts with equal number of QRS complexes which thus resulted in a total of 118 recordings to be used for further analysis.

The ECG data was analyzed by both conventional time-domain analysis[24] and a new method for individual lead analysis. In the conventional method, bidirectional bandpass filtering of the orthogonal X, Y and Z leads is done by using a fourth order Butterworth filter. The three bandpass filtered leads are then combined into a VM from which a set of measurements was computed. All measurements were made with reference to the QRS endpoint found from a

backwards search in the VM. The QRS endpoint was identified by a threshold taken as the mean noise level plus three standard deviations (SD); the noise level was measured in the VM in an interval from 100 to 140 ms after the endpoint of the unfiltered QRS complex.

A new time-domain method was employed for analysis of late potentials in individual leads.[25] The primary purpose of this method was to reliably identify the endpoint of electrical activity which is the key parameter when analyzing late potentials. The method is based on a statistical model in which certain properties of the ECG are further exploited, e.g. that cardiac activity is considered as repetitive from beat to beat while the noise is not. The amplitude of late potentials is an unknown quantity in the model which is estimated by means of a maximum likelihood (ML) procedure[26]. Based on the assumption that the ECG samples are normally distributed, this procedure chooses the most likely values of the amplitude. The resulting amplitude function was then used for finding the endpoint by means of template matching. The same shape of the template was used for all patients except for an amplitude scale factor which depends on the noise level.

Bidirectional highpass filtering was used to suppress unwanted, low-frequency components which may correlate from beat to beat, i.e. the ST-T segment. Each individual beat was subjected to highpass filtering since the noise level was computed across the ensemble of filtered beats rather than from the filtered beat average; the noise level was directly related to the standard deviation of the samples across the beat ensemble.

A few important differences between the ML method and the conventional time domain analysis should be pointed out. Although signal averaging was used for both techniques, the new method also contains information on how well the cardiac activity correlates across the ensemble of beats. This was done by splitting the ensemble into two groups. The corresponding beat subaverages were then computed using beats with either odd or even indices. Another difference is that by determining the noise level from the ensemble of beats, the problem of finding a properly located interval in the VM for noise measurement is obviated.

Filtering. The analysis of the SAECG was performed with two different highpass filter cut-off frequencies, 40 and 60 Hz. The lowpass cut-off frequency was set to 250 Hz for the VM method while no lowpass filtering was employed when the data was analyzed by the ML method. The use of lowpass filtering in combination with the ML method is avoided in order to comply with certain assumptions of the underlying statistical model.[32]

Measurements. The QRS duration (QRSD) was computed from the highpass filtered signal using the above endpoint definitions. The QRS onset was determined by a projection algorithm operating on the unfiltered beat average[27]. Since the ML method processes each individual lead, the QRSD was taken from the lead with the longest duration and was denoted $QRSD_{ML}$ (i.e. the

latest endpoint of the leads was used for computing the "over-all" QRS duration). The QRS duration obtained from the VM was denoted $QRSD_{VM}$.

The root-mean-square voltage of the terminal 40 ms (RMS40) of the QRS was obtained from the VM for both methods although related to different endpoints and were denoted as $RMS40_{ML}$ and $RMS40_{VM}$, respectively. Beat subaveraging was used only for endpoint determination in the ML method. The third common measurement is the duration of low amplitude signals < 40 μV of the terminal QRS (LAS40); this measurements was not investigated in this study because it is well-known that LAS40 is highly correlated to QRSD.

Noise level. The noise level was defined as the lowest mean voltage during a 50 ms interval from the VM, measured from the endpoint in the unfiltered QRS complex[27] and 250 ms beyond.

Definitions. The SAECG was categorized as abnormal if both QRSD and RMS40 were larger than 120 ms and less than 20 μV, respectively. The diagnostic reproducibility was established only for the 40 Hz filter due to the lack of corresponding diagnostic criteria for late potentials using 60 Hz filtering.

Statistical analysis

The difference in an SAECG variable between recordings was calculated as the absolute value of the measurement difference between two SAECG recordings for each patient. The data was expressed as mean ± one SD. The percentage change of the measurement between two recordings was calculated as the absolute difference between recordings divided by mean value of the measurement. Paired Student´s *t* test was used to compare measurements which were acquired during two recordings using different methods and filter settings. Statistical significance was defined as a *p* value less than 0.05.

RESULTS

Prevalence of late potentials

Each recording processed with the 40 Hz filter was classified as normal or abnormal. Using the ML method, 33 patients were classified as having a normal SAECG of the first recording, which remained normal in the second recording. Of the 26 patients with an initially abnormal SAECG one became normal during the second recording. In this particular case, the difference in QRSD between the two recordings was 44 ms, as opposed to 29 ms when the VM analysis was used (Fig. 1).

For the VM method, one of 34 patients with an initially normal SAECG became abnormal during the second recording and of the 25 initially abnormal recordings two became normal. Thus, using the VM method 3 out of 59 patients showed inconsistent results in the two recordings. For the patients with inconsistent results the mean difference in QRSD between two recordings was 36 ± 19.9 ms using VM analysis while using the ML method this difference was only 4.7 ± 3.2 ms.

Using the new method, one patient had late potentials in both recordings, while the SAECG was classified as normal in both recordings using VM analysis. Late potentials in individual leads can be easily lost or ventricular activity can be underestimated during combination of the individual leads into the VM as shown in Fig. 2 and 3.

Reproducibility in SAECG parameters.

Filtered QRS duration.

The mean values of the QRSD, RMS40 and noise level using the VM and the ML method with individual lead analysis during two recordings are shown in Table 1. Using 60 Hz filtering, the $QRSD_{ML}$ was significantly longer than the $QRSD_{VM}$ in both recordings ($p = 0.0001$ for the first recording and $p = 0.0029$ for the second recording). Similar results were found for the 40 Hz filter: the $QRSD_{ML}$ remained significanly longer in comparison to $QRSD_{VM}$ in recording one ($p = 0.0058$) while a trend was noted towards a longer $QRSD_{ML}$ in the second ($p = 0.058$).

The $QRSD_{ML}$ did not differ significantly between the first and the second recording for both filter settings (Table 1); the same relation applied to the $QRSD_{VM}$. The mean values of the absolute differences for all variables between the recordings are listed in Table 2. The mean absolute difference of the $QRSD_{ML}$ between the recordings was significantly lower than that of the $QRSD_{VM}$ when the 60 Hz filter was used (2.1 ± 2.2 ms vs 5.9 ± 10.2 ms, $p = 0.0038$). Using 40 Hz filtering this difference was not statistically significant, although there was a trend towards a smaller mean difference using the ML method compared to the VM method (3.7 ± 7.1 ms vs 6.9 ± 13.0 ms, $p = 0.087$). The mean percentage differences of the measurements between the two recordings are listed in Table 3. Using 60 Hz filtering the mean percentage change was significantly smaller for $QRSD_{ML}$ than that for $QRSD_{VM}$ (1.8 ± 2.1% vs 5.2 ± 9.1

%, respectively, $p = 0.0041$). Using 40 Hz filtering the mean percentage change seemed smaller for $QRSD_{ML}$, but the difference was not statistically significant (3.1 ± 5.9 % vs 5.7 ± 9.9 %, $p = 0.058$).

The mean absolute differences of the $QRSD_{ML}$ for individual X, Y and Z leads are listed in Table 2. The mean absolute difference of the QRSD was largest for the X lead and smallest for the Z lead for both filtering settings (4.6 ± 8.3 ms vs 2.1 ± 2.7 ms, for 60 Hz, p =0.034; 4.7 ± 10.4 ms vs 2.3 ± 2.6 ms, for 60 Hz, p = NS). The QRSD was longest in lead Z in 68 out of 118 recordings for 40 Hz filtering and in 59 out of 118 recordings using 60 Hz filtering.

Root-mean-square voltage of the terminal 40 ms

The mean value of the $RMS40_{ML}$ was significantly lower than the $RMS40_{VM}$ for both recordings and both filter settings ($p < 0.05$ for 40 Hz and $p < 0.001$ for 60 Hz) (Table 1).

The $RMS40_{VM}$ and $RMS40_{ML}$ did not differ significantly between the two recordings using either 40 or 60 Hz filtering (Table 1). The mean absolute difference of $RMS40_{ML}$ seemed smaller when compared to that of $RMS40_{VM}$ but did not reach statistical significance ($p = 0.065$ for 40 Hz, $p = 0.057$ for 60 Hz) (Table 2). The mean percentage change between recordings was significantly higher for RMS40 than for QRSD for both methods ($p = 0.0001$) (Table 3). Using 40 Hz filtering, the mean percentage change was 22 ± 34% for $RMS40_{ML}$ and 34 ± 48% for $RMS40_{VM}$ ($p = 0.066$). Using 60 Hz filtering the mean percentage change was still rather large but was significantly smaller for $RMS40_{ML}$ than for $RMS40_{VM}$ (12 ± 13% vs 25 ± 38%, $p = 0.012$).

The magnitude of discordant measurements between two recordings using different methods and filter settings are shown graphically in Figs. 4 and 5. The new method thus seem to provide a better reproducibility for QRSD as well as for RMS40. Both type of measurements were more reproducible when the higher filter cut-off frequency at 60 Hz was used.

Noise level.

The noise levels for both filter settings were less than 0.8 μV and did not differ significantly from recording to recording (Table 1).

DISCUSSION

Several studies regarding the reproducibility of the measurements from SAECG have been published in which parameters defined either in time, frequency or time/frequency domain have been investigated. The consensus of the studies is that time domain parameters are more reproducible than parameters which involve a spectral description.[16,17,28,29] In particular, the QRSD is the most reproducible time domain parameter. However, certain limitations of late potential measurements from the VM have been pointed out from a theoretical perspective[30] as well as from reproducibility considerations[31] suggesting that the use of individual lead analysis is preferable. In a recent paper, Lander et al[18] conclude that an increase of up to 10% in sensitivity of the SAECG could be achieved by employing individual lead analysis for identification of patients with ventricular tachycardia. The main objective of the present study was therefore to investigate if the use of individual lead analysis produces reproducible measurements when computed by a new time domain method.

Another important aspect in the assessment of a new method is how well it performs in terms of diagnostic criteria (i.e. sensitivity and specificity). Such a study, which still remains to be done, requires a considerably larger material and a more homogenous population than the one used here. When considering reproducibility only, however, it is more important to study a population which includes a large number of patients with various changes in late potentials rather than a homogenous population, e.g. only patients with sustained ventricular tachycardia or only post myocardial infarction patients.

Reproducibility of QRSD. Our study confirms earlier findings that the QRSD is the most reproducible measurement although individual lead analysis was employed in the present study (Table 3). The mean absolute difference of $QRSD_{ML}$ was significantly smaller than that of $QRSD_{VM}$; the difference was almost three times smaller when using the 60 Hz filter. Even if one would increase the scattering of $QRSD_{ML}$ by 60% the reproducibility would remain significantly better than that of the VM method. An example of this behaviour was evident in the analysis of the QRS endpoints which determined by the VM method differed considerably more from recording to recording than those of the ML method (26 ms and 4 ms, respectively) (Fig. 3) .

Engel et al.[16] presented similar results using the VM and 40 Hz filtering but found a much lower variability than in the present paper (3.0±2.4 versus 6.9±13 ms). Their material consisted, however, of 18 normal subjects for which the presence of late potentials was unlikely and accordingly the endpoint determination was a considerably less critical task.

Sager et al.[14] studied patients primarily with coronary artery disease of which 50% had late potentials but found surprisingly an even smaller mean absolute difference in QRSD (2.5±2.7 ms; 40 Hz) than that obtained by Engel et al.[16] In that study, however, 7% of the

patients were excluded due to the misdefinition of the onset or end of the QRS in the computer analysis,[14] which may have influenced the results. In our study it was found that the exclusion of a similar percentage of cases with a difference in QRSD between recordings larger than 20 ms essentially halved the over-all variability to 3.0±3.7 ms using the VM method (40Hz; 4 cases excluded) and 3.9±5.6 ms (60Hz; 6 cases excluded). It is therefore likely that the reproducibility of the VM method used in this study compares reasonably well with that used by Sager et al.[14]. Similar to the study by Malik et al[17], we did not, however, correct or exclude data on the basis of visual judgement since the object was to test the reproducibility of a standardized SAECG analysis.

The mean $QRSD_{ML}$ was significantly longer than the mean $QRSD_{VM}$ which applied for both filter settings. It is therefore obvious that late potentials were more frequently found by the ML than by the VM method, while at the same time a higher reproducibility was retained. In another comparative study, considerably longer QRSDs were obtained with the ML method than with the VM method when 100 beats were used for averaging.[25] Using 400 beats, the QRSDs obtained by the VM method approached those of the ML method which still used averages with 100 beats. This result implies that a shorter data acquistion time should be required for the ML method.

Selecting the longest QRSD from individual leads was not always equivalent to using identical lead in successive recordings. In 10 cases using 40 Hz filter and in 16 cases using 60 Hz filter the longest QRSD appeared in different leads during first and second recording. Despite such lead switching, the best reproducibility for the 60 Hz filter was found in the global $QRSD_{ML}$ and not in individual leads. The only lead that had the same mean difference of the QRSD between recordings as the $QRSD_{ML}$ was the Z lead. Changes in selected lead from successive recordings could be partly explained by the fact that the noise level differ considerably from lead to lead[23]; it has been found that the X and Y leads, in general, have a higher noise level than the Z lead does (it is thus not surprising that the Z lead was found to be the most reproducible individual lead). Biological variability is another possible explanation which is expressed as slight changes in the vector direction of the late potentials itself due to the electrical inhomogeneities of the arrhythmogenic substrate.

Reproducibility in RMS40. The mean percentage change of the QRSD for both methods and for both filter settings was significantly smaller than that for RMS40. The higher variability of RMS40 is probably related to the large variation of amplitude within the terminal interval. It is obvious that small changes in the QRS endpoint may significantly affect the samples included for analysis. The variability is further pronounced by the squaring operation in the RMS definition. The mean absolute difference between recordings and the mean percentage change were rather large and did not differ significantly between the methods. Although the

reproducibility of RMS40 was rather poor for both methods, it was slightly improved using an endpoint from the longest QRSD in individual leads with 60 Hz filtering.

Algorithmic aspects. The use of the VM is a well-established technique in commercial devices for SAECG analysis. Different manufacturers have their own proprietary schemes for endpoint determination in the VM which may differ from the one employed in this paper. It can rightfully be argued that such proprietary schemes may perform better than did the present VM method and as a result the significance of the present study appears very restricted. It is important to realize, however, that the performance of any such methods is limited by the signal-to-noise ratio of the VM.[25] This property was illustrated by Fig. 2 in which highpass filtered averaged beats were shown together with the corresponding VM. In that case, late potentials were visible primarily in the X lead while the activity in the remaining leads terminated earlier. It is apparently difficult to determine from the VM at what point in time the late potential activity terminates.

The effects of highpass filtering at various cut-off frequencies using the VM method have been thoroughly investigated.[32,33] It has been established that a higher cut-off frequency, e.g. 80 Hz, results in a higher sensitivity in detecting patients with ventricular tachycardia while a lower cut-off frequency, e.g. 40 Hz, results in a better specificity.[32,34] The present study showed that both $QRSD_{ML}$ and $RMS40_{ML}$ were more reproducible when using a highpass filter with cut-off frequency at 60 Hz instead of 40 Hz. Figure 1 shows an example in which the reproducibility is poor for both methods. In the lower panel, the slowly changing low amplitude waveform (with the possible interpretation as late potentials) starting after sample 120 is not detected because of the occurrence of a "silent" segment prior to the waveform. For 60 Hz filtering (not shown), the waveform completely disappears and only negligible variability is found in the endpoint for the methods. Using methods involving information about ensemble correlation, such as the ML method, it seems that filters with higher cut-off frequencies are required than those used with VM methods. Reminiscences of the ST-T segment after filtering may otherwise be detected as late potentials.

A disadvantage with the ML method is its computational complexity which with the present computer technology in ECG devices is prohibitive.The computational bulk is the highpass filtering needed for each individual beat and the computation of the amplitude function used for endpoint determination. The development of efficient implementations as well as the ever-increasing speed of computers will hopefully pave the way for use of the ML method in a clinical setting.

Statistical analysis. In some reproducibility studies, the results were expressed using Pearson's correlation coefficient.[9,15,28,35] However, that method merely indicates that the entire measurement of each subject is better associated with the measurement of the same subject during next recording than within the total study group. Moreover, correlation

coefficients do not give any information about within-subject variability.[36,37] Thus, reproducibility in our study was expressed in terms of the absolute value of the measurement difference between recordings and percentage of a change of the measurement.

CONCLUSIONS.

The use of an ML method for SAECG analysis resulted in significantly longer QRSD, thus indicating a possibility how to improve the sensitivity of SAECG. Moreover, the new method produced measurements which were more reproducible in comparison to the conventional time domain VM method.

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LEGENDS TO FIGURES

Fig. 1 An example of poor reproducibility in endpoint determination. The highpass filtered beat subaverages (40 Hz) are plotted for the X, Y and Z leads together with the VM. Endpoints in individual leads and VM endpoint are shown by vertical bars. The upper and lower panels show the first and second recording, respectively.

Fig. 2 Occurrence of late potentials in a single lead (X lead). The highpass filtered beat subaverages (60 Hz) are plotted for the X, Y and Z leads together with the VM. Endpoints in individual leads and VM endpoint are shown by vertical bars. The upper and lower panels show the first and second recording, respectively.

Fig. 3 Occurrence of late potentials with longest duration in the Y lead. The highpass filtered beat subaverages (60 Hz) are plotted for the X, Y and Z leads together with the VM. Endpoints in individual leads and VM endpoint are shown by vertical bars. The upper and lower panels show the first and second recording, respectively.

Fig. 4 Scatter plot of $QRSD_{ML}$ and $QRSD_{VM}$. Individual values of the difference between two recordings are plotted against values of the average of two recordings using (a) 40 and (b) 60 Hz filtering.

Fig. 5 Scatter plot of $RMS40_{ML}$ and $RMS40_{VM}$. Individual values of the difference between two recordings are plotted against values of the average of two recordings using (a) 40 and (b) 60 Hz filtering.

TABLE 1. Mean values of the QRSD, RMS40 and the noise level using the VM and ML methods during two recordings (comparisons were made between the two methods, results are expressed as mean±SD, °- p = NS, *- $p < 0.05$; Rec - recording)

	Filter	$QRSD_{ML}$ (ms)	$QRSD_{VM}$ (ms)	RMS_{VM} (μV)	RMS_{ML} (μV)	Noise (μV)
Rec 1	40 Hz	121±32	116±27*	14.6±15.6	12.7±14.4*	0.4±0.2
	60 Hz	117±32	112±27*	10.1±8.9	8.6±8.5*	0.3±0.1
Rec 2	40 Hz	119±31	115±29°	15.7±15.7	13.0±14.2*	0.4±0.3
	60 Hz	117±31	112±28*	11.0±9.9	8.3±8.3*	0.3±0.1

TABLE 2. Mean values of the absolute differences of the QRSD, RMS40 and the noise level between two recordings (comparisons were made between the two methods, results are expressed as mean±SD; Δ-absolute difference)

Δ	40 Hz	p value	60 Hz	p value
$QRSD_{VM}$ (ms)	6.9 ±13.0		5.9±10.3	
$QRSD_{ML}$ (ms)	3.8±7.1	0.087	2.1±2.2	0.0038
QRSD X (ms)	4.7±10.4	0.3	4.6±8.3	0.397
QRSD Y (ms)	2.8±2.6	0.023	3.5±3.3	0.077
QRSD Z (ms)	2.3±2.6	0.009	2.1±2.7	0.0045
RMS_{VM} (μV)	3.5±5.9		2.0±4.9	
RMS_{ML} (μV)	1.9±2.9	0.065	0.8±1.0	0.058
Noise (μV)	0.1±0.1		0.06±0.6	

TABLE 3. Mean percentage change of the QRSD and RMS40 between recordings (results are expressed as mean±SD; % - percentage change of the measurement)

	40 Hz % change	*p* value	60 Hz % change	*p* value
$QRSD_{VM}$	5.7±9.9		5.1±9.1	
$QRSD_{ML}$	3.1±5.9	0.058	1.7±2.1	0.0041
RMS_{VM}	34.0±47.8		25.3±38.8	
RMS_{ML}	21.6±34.7	0.0661	11.7±13	0.012