The Effects of 40 Hz Low-pass Filtering on the Magnitude of the Spatial Ventricular Gradient

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Abstract

Changes in the magnitude of the spatial ventricular gradient (MSVG) have been associated with an increased risk of ventricular arrhythmias. This association makes the MSVG a potentially attractive parameter for ECG monitoring applications.

The MSVG is typically obtained using 150 Hz low-pass filtered resting ECGs. However, monitoring applications typically utilise upon 40 Hz low-pass filtered ECG data. The extend to which the utilization of 40 Hz low-pass monitoring ECG filters over the commonly used 150 Hz low-pass resting ECG filters does affect the MSVG value has not previously been reported.

The aim of this research was to quantify the differences between MSVG values computed using 40 Hz low-pass filtered ECG data (MSVG40) and 150 Hz low-pass filtered ECG data (MSVG150). The differences between the MSVG40 and the MSVG150 were quantified as systematic error (mean difference) and random error (span of Bland-Altman 95% limits of agreement) using a study population of 726 subjects. The systematic error was found to be 0.013 mV ms [95% confidence interval: 0.008 mV ms to 0.018 mV ms]. The random error was quantified as 0.282 mV ms [95% confidence interval: 0.266 mV ms to 0.298 mV ms].

Our findings suggest that it is possible to record accurate MSVG values in monitoring applications that require the utilization of 40 Hz low-pass filtered ECG data.

1. Introduction

The spatial ventricular gradient (SVG) can be computed from the Frank vectorcardiogram (VCG) [1]. The SVG has previously been described [2] as a measure of the ventricular action potential heterogeneity. More precisely, the SVG has been described to reflect the heterogeneity in the ventricular action potential duration as well the heterogeneity of the ventricular action potential morphology. An increase in ventricular action potential heterogeneity has been reported to be associated with an increased risk of ventricular arrhythmias [3] and previous research [4] has identified an association between the magnitude of the SVG (MSVG) and the risk of ventricular arrhythmias. The association between the MSVG and the risk of ventricular arrhythmias can be of potential interest in applications that require the utilization of the monitoring electrocardiogram (ECG).

A requirement for the determination of the MSVG is the availability of the Frank VCG. Modern vectorcardiography commonly derives the Frank VCG from the resting standard 12-lead ECG [5], which is recorded using distal limb-electrodes that are placed at the ankles and wrists of the patient. This approach for the determination of the Frank VCG is not suitable for monitoring applications as the distal limb leads of the standard 12-lead ECG are susceptible to motion artifacts.

Recent efforts have focused on overcoming the lead system related barriers for the recording of the Frank VCG in monitoring applications. This has lead to the development of a number of different linear electrocardiographic lead transformation matrices [6], [7]. These transformation matrices allow for the derivation of the Frank VCG from different monitoring compatible lead sets. More precisely, linear ECG lead transformations utilize (1) to derive or estimate the three leads of the Frank VCG using a number of recorded basis leads.

$$\boldsymbol{VCG} = \boldsymbol{A} \cdot \begin{bmatrix} \boldsymbol{B_{L1}} \\ \vdots \\ \boldsymbol{B_{LM}} \end{bmatrix} \quad (1)$$

Where $\boldsymbol{VCG}$ is a $3\times N$ matrix containing N sample values for each of the three Frank leads, $\boldsymbol{B_{L1}}$ to $\boldsymbol{B_{LM}}$ denote $1\times N$ vectors containing N sample values for each of the $\boldsymbol{M}$ basis leads and $\boldsymbol{A}$ is the $3\times M$ transformation matrix.

The recent availability of these monitoring compatible transformation matrices has extended the usability of the

MSVG from resting to monitoring applications.

Previous studies have quantified the transformation matrix related estimation errors of different transformation matrices. For example, a performance comparison of the Kors and the Guldenring matrix found similar estimation error levels in Frank VCGs that were derived using the resting 12-lead ECG and the monitoring compatible Mason-Likar (ML) 12-lead ECG respectively [7], [8].

While the transformation matrix related estimation errors that are associated with the utilization of monitoring compatible lead sets have been quantified, the influence of the different filter strategies used for recording monitoring and resting ECGs on the value of the MSVG has not fully been investigated. More precisely, the influence of the differences in the minimum high-frequency cutoff requirement of 150 Hz for resting [9] and 40 Hz for monitoring ECGs [10] on the value of the MSVG has, to the best of our knowledge, not previously been reported in the literature.

The aim of this research is twofold. First, to quantify the difference between MSVG values that are obtained from 40 Hz and 150 Hz filtered Frank VCGs. Second, to compare this difference to the transformation matrix related estimation errors that are made when computing the MSVG from the ML derived Frank VCG.

2. Material and methods

2.1. Study population

We base our research on a study population of 726 subjects. The study population is composed of 229 normal subjects, 265 subjects with myocardial infarction and 232 subjects with left ventricular hypertrophy.

2.2. BSPM data

This research was conducted using secondary body surface potential map (BSPM) data. A total of 120 ECG leads were recorded for each BSPM. All of the 120 leads were recorded in reference to the Wilson central terminal. Three of the 120 BSPM leads were recorded from electrodes placed on the right and left wrist and the left ankle (VR, VL and VF respectively). The remaining 117 leads were recorded from 81 anterior and 36 posterior recording sites. A total of 15 seconds of continuous ECG data was recorded for the 120 ECG leads using a sample rate of 500 samples per second. One representative average P-QRS-T complex was computed for each lead using the 15 seconds of continuous ECG data. A comprehensive description of the BSPM data and the recording procedure is provided in [11].

The body surface potentials at anatomic locations that correspond to the 352 nodes of the Dalhousie torso [12] were calculated for each BSPM. This was achieved by applying a previously reported Laplacian 3D interpolation procedure [13] to the 117 thoracic BSPM leads.

2.3. Derivation of the Frank VCG

The body surface potentials at the A, C, E, F, H, I and M electrode locations of the Frank lead system were extracted from each of the 726 interpolated BSPMs.

Body surface potentials at the I and M electrode locations of the Frank lead system were not directly located at one of the 352 nodes defined by the Dalhousie torso. These body surface potentials were determined through linear interpolation of body surface potentials located at proximal nodes of the Dalhousie torso.

The body surface potentials at the A, C, E, F, H, I and M electrode locations were used to derive the three leads of the Frank VCG using (2).

$$\boldsymbol{VCG} = \begin{bmatrix} \boldsymbol{X} \\ \boldsymbol{Y} \\ \boldsymbol{Z} \end{bmatrix} = \boldsymbol{A} \cdot \begin{bmatrix} \boldsymbol{\varphi_A} \\ \vdots \\ \boldsymbol{\varphi_M} \end{bmatrix}. \quad (2)$$

Where $\boldsymbol{\varphi_A}$, $\boldsymbol{\varphi_C}$, $\boldsymbol{\varphi_E}$, $\boldsymbol{\varphi_F}$, $\boldsymbol{\varphi_H}$, $\boldsymbol{\varphi_I}$, and $\boldsymbol{\varphi_M}$ are $1\times N$ vectors that contain N sample values of potentials at the Frank electrode locations A to M respectively, N denotes the number of samples in the average P-QRST complex, $\boldsymbol{A}$ is a 3×7 matrix of published coefficients [14] that allow for a derivation of the Frank VCG using the potentials $\boldsymbol{\varphi_A}$ to $\boldsymbol{\varphi_M}$, and $\boldsymbol{VCG}$ is a $3\times N$ matrix containing N sample values of the Frank VCG, the $1\times N$ vectors $\boldsymbol{X}$, $\boldsymbol{Y}$ and $\boldsymbol{Z}$ contain N sample values of the three Frank leads X, Y and Z respectively.

2.4. Low-pass filtering of the Frank VCG

One 40 Hz and one 150 Hz low-pass filtered Frank VCG was generated for each subject in the study population. This was achieved by applying one 40 Hz and one 150 Hz low-pass filter to each of the 726 Frank VCGs that were extracted from the interpolated BSPM data.

Filtering was performed using two different phase-linearized infinite impulse response (IIR) digital low-pass filters. The utilized IIR filters were based upon 6^{th} order Butterworth filters with corner frequencies located at 40 Hz and 150 Hz. These filters were cascaded with group-delay equalizers to yield an approximately linear phase response to avoid filtering artifacts. Each group-delay equalizer was implemented as IIR allpass filter and designed using the method described in [15]. The filter characteristics (passband average group-delay [16]; passband group-delay deviation [16]) of the 40 Hz and the 150 Hz phase-linearized low-pass filter were quantified to as (17.73 samples; 0.90 samples) and (10.90 samples; 0.29 samples) respectively.

2.5. Determination of the magnitude of the spatial ventricular gradient

The MSVG for each of the low-pass filtered Frank VCGs was determined using (3) and (4).

$$MSVG40^i = \Delta T \cdot \sum_{n=QRS_{ON}}^{T_{END}} \boldsymbol{VCG40^i}(n). \quad (3)$$

$$MSVG150^i = \Delta T \cdot \sum_{n=QRS_{ON}}^{T_{END}} \boldsymbol{VCG150^i}(n). \quad (4)$$

Where $\boldsymbol{VCG40}$ is a $3\times N$ matrix containing N sample values of each of the three 40 Hz low-pass filtered Frank VCG leads, $\boldsymbol{VCG150}$ denotes a $3\times N$ matrix containing N sample values of each of the three 150 Hz low-pass filtered Frank VCG leads, QRS_{ON} is the sample index of the QRS onset, T_{END} denotes the sample index associated with the end of the T wave, ΔT denotes the sample interval used when recording the BSPM data, $MSVG40$ and $MSVG150$ refer to MSVG values that are determined using 40 Hz and 150 Hz low-pass filtered Frank VCG data respectively, $i \in \{1, \dots, 726\}$ is an index variable that is used to indicate the MSVG value that is associated with the i-th subject in the study population.

2.6. Quantification of the effect of 40 Hz low-pass filtering on the magnitude of the spatial ventricular gradient

The effect of the 40 Hz low-pass filter on the value of the MSVG was quantified using a multistep procedure. First, the differences between the $MSVG40$ values and the $MSVG150$ values were calculated as detailed in (5).

$$\Delta\boldsymbol{MSVG} = \boldsymbol{MSVG40} - \boldsymbol{MSVG150}. \quad (5)$$

Where $\boldsymbol{MSVG40}$ and $\boldsymbol{MSVG150}$ are vectors that contain the $MSVG40^i$ and the $MSVG150^i$ values of all subjects in the study population and $\Delta\boldsymbol{MSVG}$ is a vector that contains the differences between the $MSVG40^i$ and the $MSVG150^i$ values of all subjects in the study population.

Second, the systematic error of the differences between the $MSVG40^i$ and the $MSVG150^i$ values was quantified by computing the mean [95% confidence intervals (CI)] and the median [95% CI] of the values in $\Delta\boldsymbol{MSVG}$.

Third the random error component of the differences between the $MSVG40^i$ and the $MSVG150^i$ values was quantified. This was achieved by calculating the span of the Bland-Altman 95% limits of agreement as detailed in (6).

$$\text{RandomError} = 2 \cdot 1.96 \cdot std(\Delta\boldsymbol{MSVG}). \quad (6)$$

Where $std(\cdot)$ denotes the standard deviation and $\Delta\boldsymbol{MSVG}$ is as defined in (5).

The random error component was, in addition to the Bland-Altman 95% limits of agreement also quantified as the interquartile range [95% CI] of the values in $\Delta\boldsymbol{MSVG}$.

The 95% confidence intervals for the median $\Delta\boldsymbol{MSVG}$ value, for the span of the Bland-Altman 95% limits of agreement and the for interquartile range of the values in $\Delta\boldsymbol{MSVG}$ was determined using bootstrapping. More precisely, 10000 bootstrap replicates were used to calculate the bootstrapped bias-corrected and accelerated 95% confidence interval (95% CI) [17] for each of these parameters.

3. Results

The values of the systematic error and the random error component that were identified in this research are provided in Table 1.

Table 1. Differences between the MSVG values obtained from 40 Hz low-pass and 150 Hz low-pass filtered Frank VCG data.

Parameter	Parameter value	95% Confidence Interval
Random error		
BA limits of agreement[a]	0.282	[0.266 to 0.298]
interquartile range[b]	0.091	[0.085 to 0.102]
Systematic error		
mean[c]	0.013	[0.008 to 0.018]
median[d]	0.015	[0.009 to 0.020]

Notes. [a]Bland-Altman 95% limits of agreement and [b]interquartile range of the MSVG differences in $\Delta\boldsymbol{MSVG}$, [c]mean and [d]median of the MSVG differences in $\Delta\boldsymbol{MSVG}$. Parameter values and 95% confidence intervals are given in mV ms.

4. Discussion

In this research we have quantified the changes in the value of the MSVG due to the application of a 40 Hz low-pass ECG monitoring filter to directly recorded Frank VCG data.

Our findings suggest that the utilisation of a 40 Hz low-pass ECG monitoring filter on the recorded Frank VCG is associated with a relatively small systematic error of 0.013 mV ms [95% CI: 0.008 mV ms to 0.018 mV ms. It is possible reduce the influence of this already small systematic error by subtracting the point estimate of this error component (0.013 mV ms) from MSVG values that are computed using 40 Hz low-pass filtered VCG data. In addition, the 40 Hz low-pass ECG monitoring filter was found to introduce a random error component of 0.282 mV ms [95% CI: 0.266 mV ms to 0.298 mV ms] to the MSVG values.

Modern vectorcardiography typically relies upon derived VCGs that are obtained through the utilization of

linear ECG lead transformations rather then directly recorded Frank VCGs. The two lead transformation matrices that are commonly used for deriving the Frank VCG from the standard 12-lead ECG and the ML 12-lead ECG are the Kors [18] and the Guldenring matrix [7] respectively. Previous research has quantified the Guldenring matrix related estimation error when computing the MSVG from the ML 12-lead ECG as (systematic error: -3.00 mV ms; random error: 30.17 mV ms) [19]. This lead transformation related estimation error is large compared to the influence of the 40 Hz low-pass ECG monitoring filter on the value of the MSVG. From this we conclude that the utilization of 40 Hz low-pass filtered ECG data is no obstacle for the determination of the MSVG in ECG monitoring applications.

5. Conclusion

In this paper we have reported on the effects of a 40 Hz low-pass ECG monitoring filter on the MSVG. Our findings suggest that linear-phase (or alternatively zero-phase) ECG monitoring filters with a high-frequency cutoff of 40 Hz do only introduce minor changes to the MSVG.

Based on our findings we conclude that it is possible to record the MSVG in ECG monitoring applications that require the utilization of 40 Hz low-pass filters.

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