The Effects of 40 Hz Low-pass Filtering on the Spatial QRS-T Angle

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Abstract

The spatial QRS-T angle (SA) is a vectorcardiographic (VCG) parameter that has been identified as a marker for changes in the ventricular depolarization and repolarization sequence. The SA is defined as the angle subtended by the mean QRS-vector and the mean T-vector of the VCG. The SA is typically obtained from VCG data that is derived from the resting 12-lead electrocardiogram (ECG). Resting 12-lead ECG data is commonly recorded using a low-pass filter with a cutoff frequency of 150 Hz. The ability of the SA to quantify changes in the ventricular depolarization and repolarization sequence make the SA potentially attractive in a number of different 12-lead ECG monitoring applications. However, the 12-lead ECG data that is obtained in such monitoring applications is typically recorded using a low-pass filter cutoff frequency of 40 Hz. The aim of this research was to quantify the differences between the SA computed using 40 Hz low-pass filtered ECG data (SA40) and the SA computed using 150 Hz low-pass filtered ECG data (SA150). We assessed the difference between the SA40 and the SA150 using a study population of 726 subjects. The differences between the SA40 and the SA150 were quantified as systematic error (mean difference) and random error (span of Bland-Altman 95% limits of agreement). The systematic error between the SA40 and the SA150 was found to be -0.126° [95% confidence interval: -0.146° to -0.107°]. *The random error was quantified* 1.045° [95% confidence interval: 0.917° to 1.189°]. *The findings of this research suggest that it is possible to accurately determine the value of the SA when using 40 Hz low-pass filtered ECG data. This finding indicates that it is possible to record the SA in applications that require the utilization of 40 Hz low-pass ECG monitoring filters.*

1. Introduction

The relationship between the ventricular depolarization and ventricular repolarization can be quantified using the spatial QRS-T angle (SA). The SA is sensitive to changes in the depolarization sequence as well as to changes in the action potential duration. The SA is a parameter that is computed from the Frank vectorcardiogram (VCG) [1]. However, the Frank VCG is not typically recorded in modern day clinical practice. The SA is therefore frequently determined using estimated or derived VCG data. The derived VCG data is typically obtained though the utilization of linear electrocardiographic lead transformation matrices that are applied to the 12-lead electrocardiogram (ECG). None of the established electrocardiographic lead transformation matrices (such as for example the Kors matrix [2] or the inverse Dower matrix [3]) has been optimized for the derivation of the Frank VCG using monitoring compatible electrocardiographic lead sets. Previous research on the utility of the SA was therefore confined to clinical applications that are compatible with the recording of diagnostic (resting) 12-lead ECG data. However, the determination of the SA in monitoring applications is of potential clinical interest. This is because the SA can identify an abnormal relationship between ventricular depolarization and ventricular repolarization. This is of interest as repolarization and depolarization abnormalities are risk factors for the development ventricular arrhythmias [4]. Recent efforts have focused on overcoming the lead system related barriers for the utilization of the SA in monitoring applications. This has lead to the development of different linear electrocardiographic lead transformation matrices that allow for the derivation of the SA using monitoring compatible electrocardiographic lead sets. The Guldenring matrix [5-7] for example allows for the derivation of the SA from the monitoring compatible Mason-Likar (ML) 12-lead ECG [8]. In addition, linear electrocardiographic lead transformation matrices, that can be used to derive the SA from all 62 different reduced lead systems that contain ML limb leads I, II and all possible combinations of precordial leads V1 to V6, have recently been developed [7]. The recent advances in the availability of monitoring compatible electrocardiographic lead transformation matrices have removed barriers for the utilization of the SA in

monitoring applications. However, monitoring ECGs and diagnostic (resting) ECGs do not only require the use of different electrocardiographic lead sets they also utilize upon different signal filter characteristics. The American Heart Association recommends that diagnostic (resting) ECGs should be recorded using signal filters with a minimum high-frequency cutoff of 150 Hz [9]. This is different to monitoring ECGs where a minimum high-frequency cutoff of 40 Hz is required [10]. It is known that the utilization of the 40 Hz high-frequency cutoff in monitoring applications is associated with a reduction of QRS amplitudes. Whether ECG monitoring filters do have an influence on the value of the SA has, to the best of our knowledge, not previously been reported in the literature. The aim of this research is to quantify the effect of the 40 Hz high-frequency cutoff, that is used in ECG monitoring filters, on the value of the SA.

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

One body surface potential map (BSPM) was recorded for each of the 726 subjects in the study population. Each BSPM used in this research contains electrocardiographic data of 120 BSPM leads. A representative average P-QRST complex was calculated for each of the 120 BSPM leads. Three of the 120 leads were recorded from electrodes placed on the right and left wrist and the left ankle (VR, VL and VF respectively). Electrodes situated at 81 anterior and 36 posterior locations were used to record 117 thoracic leads. All thoracic leads were recorded with reference to the Wilson central terminal. A comprehensive description of the BSPM data and the recording procedure can be found in [11].

2.3. Extraction of the Frank VCG data

One Frank VCG was extracted from each of the 726 BSPMs. However, some of the body surface potentials that are used by the Frank VCG were associated with electrode locations that were not covered by the thoracic electrode grid. A previously reported two-step interpolation procedure [12] was used to obtain the required body surface potentials that were not directly recorded by the thoracic electrode grid.

First, a Laplacian 3D interpolation procedure was applied to the 117 recorded thoracic leads. This was performed to obtain the body surface potentials at the locations of the 352 Dalhousie torso [13] nodes. Second, linear interpolation was used to obtain all required thoracic body surface potentials that were located between the Dalhousie torso node. The body surface potentials at the A, C, E, F, H, I and M electrode locations of the Frank lead system were extracted from the interpolated BSPM data and subsequently used to derive the Frank VCGs.

2.4. Low-pass filtering of the Frank VCG

Two low-pass filtered versions of each Frank VCG were generated. This was achieved by applying one 40 Hz and one 150 Hz low-pass filter to each lead of the 726 Frank VCGs. We subsequently utilize $\boldsymbol{VCG^{40Hz}}$ and $\boldsymbol{VCG^{150Hz}}$ to refer to 40 Hz low-pass filtered and 150 Hz low-pass filtered Frank VCGs respectively. The filtered Frank VCGs were generated using 6th order Butterworth infinite impulse response (IIR) digital low-pass filters with corner frequencies located at 40 Hz and 150 Hz. The non-linear phase responses of the low-pass filters were approximately linearized in order to avoid filter artifacts. Phase-linearization was performed by cascading one group-delay equalizer with each of the two low-pass filters. The group-delay equalizers were implemented as IIR allpass filters and designed using the method described in [14]. The group-delay characteristics of the cascaded filter structure (low-pass filter in series with the group-delay equalizer) were quantified as passband average group-delay (τ_{avg}) and passband group-delay deviation (τ_d) using (1) and (2) respectively.

$$\tau_{avg} = \frac{\tau_{max} + \tau_{min}}{2}. \quad (1)$$

$$\tau_d = \tau_{max} - \tau_{min}. \quad (2)$$

Where τ_{avg} refers to the passband average group-delay, τ_{max} is the maximal group-delay in the filter passband, τ_{min} is the minimal group-delay in the filter passband and τ_d refers to the passband group-delay deviation.

Both, the passband average group-delay as well as the passband group-delay deviation of the phase-linearized low-pass filters are detailed in Table 1.

Table 1. Passband average group-delay and passband group-delay deviation of the phase-linearized 40 Hz and 150 Hz low-pass filters.

Filter type	τ_{avg} [samples]	τ_d [samples]
40 Hz	17.73	0.90
150 Hz	10.90	0.29

2.5. Determination of the SA

The SA values were calculated using the low-pass filtered Frank VCGs as detailed in (3) to (8).

$$\boldsymbol{QRS^d} = \frac{1}{J_p - QRS_{ON}} \sum_{n=QRS_{ON}}^{J_P} \boldsymbol{VCG^d}(n). \quad (3)$$

$$\boldsymbol{T^d} = \frac{1}{T_{END} - J_P} \sum_{n=J_P}^{T_{END}} \boldsymbol{VCG^d}(n). \quad (4)$$

$$SA40 = arccos\left[\frac{\boldsymbol{QRS^{40Hz}} \cdot \boldsymbol{T^{40Hz}}}{|\boldsymbol{QRS^{40Hz}}| \cdot |\boldsymbol{T^{40Hz}}|}\right]. \quad (5)$$

$$SA150 = arccos\left[\frac{\boldsymbol{QRS^{150Hz}} \cdot \boldsymbol{T^{150Hz}}}{|\boldsymbol{QRS^{150Hz}}| \cdot |\boldsymbol{T^{150Hz}}|}\right]. \quad (6)$$

$$SA40QRS = arccos\left[\frac{\boldsymbol{QRS^{40Hz}} \cdot \boldsymbol{T^{150Hz}}}{|\boldsymbol{QRS^{40Hz}}| \cdot |\boldsymbol{T^{150Hz}}|}\right]. \quad (7)$$

$$SA40T = arccos\left[\frac{\boldsymbol{QRS^{150Hz}} \cdot \boldsymbol{T^{40Hz}}}{|\boldsymbol{QRS^{150Hz}}| \cdot |\boldsymbol{T^{40Hz}}|}\right]. \quad (8)$$

Where $\boldsymbol{QRS^d}$ is the 3×1 mean vector of ventricular depolarization, $\boldsymbol{T^d}$ denotes the 3×1 mean vector of ventricular repolarization, QRS_{ON} is the sample index of the QRS onset, J_P denotes the sample index of the J-point, T_{END} is the sample index associated with the end of the T wave, $\boldsymbol{VCG}^d$ is a $3\times N$ matrix containing N sample values of the three filtered Frank VCG leads, $d \in \{40\,Hz, 150\,Hz\}$ denotes whether a parameter was calculated based upon 40 Hz or 150 Hz low-pass filtered Frank VCGs, $SA40$ and $SA150$ refer to SA values that are determined using 40 Hz and 150 Hz low-pass filtered Frank VCGs respectively, $SA40QRS$ denotes a SA value that is calculated using the mean QRS-vector and the mean T-vector obtained from 40 Hz and 150 Hz low-pass filtered Frank VCGs respectively, $SA40T$ denotes a SA value that is calculated using the mean QRS-vector and the mean T-vector obtained from 150 Hz and 40 Hz low-pass filtered Frank VCGs respectively.

2.6. Quantification of the effect of 40 Hz low-pass filtering on the SA

The effect of the 40 Hz low-pass filter on the value of the SA was quantified. This was performed using a multistep procedure. First, the differences between the $SA40$ values and the $SA150$ values were calculated as detailed in (9).

$$\Delta\boldsymbol{SA} = \boldsymbol{SA40} - \boldsymbol{SA150}. \quad (9)$$

Where $\boldsymbol{SA40}$ and $\boldsymbol{SA150}$ are vectors that contain the $SA40$ and the $SA150$ values of all subjects in the study population and $\Delta\boldsymbol{SA}$ is a vector that contains the differences between the $SA40$ and the $SA150$ values of all subjects in the study population.

Second, the systematic and the random error component of the differences between the $SA40$ and the $SA150$ values were analyzed. The systematic error was quantified as mean [95% confidence intervals (CI)] of the elements in $\Delta\boldsymbol{SA}$. We quantified the random error using the span of the Bland-Altman 95% limits of agreement as detailed in (10).

$$\text{RandomError} = 2 \cdot 1.96 \cdot std(\Delta\boldsymbol{SA}). \quad (10)$$

Where $std(\cdot)$ denotes the standard deviation and $\Delta\boldsymbol{SA}$ is as defined in (9).

Third, the contribution of the 40 Hz low-pass filter related changes in the mean QRS-vector and the 40 Hz low-pass filter related changes of the mean T-vector to the ΔSA values was assessed. This was performed through the use of the linear model in (11).

$$\widehat{\Delta SA} = b_1 \cdot \Delta SA^{40HzQRS} + b_2 \cdot \Delta SA^{40HzT}. \quad (11)$$

$$\Delta SA^{40HzQRS} = (SA40QRS - SA150). \quad (11a)$$

$$\Delta SA^{40HzT} = (SA40T - SA150). \quad (11b)$$

Where b_1 and b_2 are the coefficients of the linear model, $SA150$ and $SA40QRS$ and $SA40T$ are as defined in (6) to (8) respectively, $\widehat{\Delta SA}$ is an estimate of the ΔSA value associated with one subject in the study population, $\Delta SA^{40HzQRS}$ denotes the contribution of the 40 Hz low-pass filter related changes in the mean QRS-vector to the ΔSA value and ΔSA^{40HzT} denotes the contribution of the 40 Hz low-pass filter related changes in the mean T-vector to the ΔSA value.

Random sampling was used to divide the study population into a training dataset (DTrain) and a testing dataset (DTest). The coefficients in (12) were developed using the Frank VCGs of the 242 subjects in DTrain and linear least squares regression. The performance of the linear model in (11) was assessed using the Frank VCGs of the 484 subjects in DTest. The performance of the linear model in (11) was quantified by the Pearson product-moment correlation coefficient and the Root-Mean-Square Difference (RMSD) between the $\widehat{\Delta SA}$ and the ΔSA values of all subjects in DTest. We used the mean absolute magnitude to quantify the scale of the predictor Variables $\Delta SA^{40HzQRS}$ and $\Delta SA^{40HzQRS}$.

3. Results

The analysis of the values in **ΔSA** found that the utilization of the 40 Hz high-frequency cutoff that is used in ECG monitoring filters is associated with a systematic error of -0.126° [95% CI: -0.146° to -0.107°]. In addition, the 40 Hz high-frequency cutoff was found to be associated with a random error component of 1.045° [95% CI: 0.917° to 1.189°]. Using least squares linear regression analysis the coefficients in (12) were found to be b_1= 1.081 [95% CI: 1.048 to 1.114] and b_2= 1.065

[95% CI: 1.001 to 1.121]. The Pearson product-moment correlation coefficient between the $\Delta\widehat{SA}$ and the ΔSA values of all subjects in DTest was found to be 0.961 [95% CI: 0.954 to 0.968] and the RMSD between the $\Delta\widehat{SA}$ and the ΔSA values of all subjects in DTest was determines as 0.095° [95% CI: 0.067° to 0.130°]. The mean absolute magnitudes of the $\Delta SA^{40HzQRS}$ values and the SA^{40HzT} values of all subjects in DTest were found to be 0.192° and 0.094° respectively.

4. Discussion

The random error component that is introduced through the utilization of a 40 Hz low-pass ECG monitoring filter was found to be 1.045° [95% CI: 0.917° to 1.189°]. The derivation of the SA from the 12-lead ECG has been reported to be associated with random error magnitudes of the order of 40° to 55° [7, 15]. This is substantially higher than what is introduced through the utilization of the 40 Hz low-pass ECG monitoring filter. Given the relatively small influence of the 40 Hz low-pass ECG monitoring filter on the SA, it is clear that $SA40$ values can, from a clinical perspective, be regarded as equivalent substitutes for $SA150$ values.

From the coefficient values of the linear model in (11) (b_1= 1.081; b_2= 1.065) and the differences in the scale of the predictor variables (mean absolute magnitudes of the $\Delta SA^{40HzQRS}$ values, 0.192°; mean absolute magnitudes of the ΔSA^{40HzT} values, 0.094°), one can identify the 40 Hz low-pass filter related changes to the mean QRS-vector as the largest contributor to the ΔSA.

5. Conclusion

This paper reported on the effects of the 40 Hz low-pass ECG monitoring filter on the SA. The 40 Hz low-pass ECG monitoring filter was found to introduce only minor changes to the SA. This finding suggests that it is possible to record the SA in applications that require the utilization of 40 Hz low-pass ECG monitoring filters.

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