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A novel method for cardiac vector velocity measurement: Evaluation in myocardial infarction

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ABSTRACT

Background and objective: Pathological alterations provoked by myocardial infarction cause slow conduction by increasing axial resistance on coupling between cells. This issue may cause abnormal patterns in the dynamics of the tip of the cardiac vector.
Methods: In this work, we have developed a method to compute the angular velocity during ventricular repolarization from Frank *XYZ* leads, using the concept of quaternion. This parameter jointly with the linear velocity obtained by differentiation and the spatial velocity reported by others during ventricular depolarization, have been combined in order to design a myocardial infarction detector (so-called index of cardiac vector velocity: ICVV) with high values of sensitivity and specificity simultaneously.
Results: The predictive power of ICVV has been tested in two groups: patients with less than 7 days after infarction, achieving 98% of sensitivity and 97% of specificity; and patients with more than 45 days after infarction, achieving 92% of sensitivity without loss of specificity. The former group is important for early detection of myocardial infarction and begins treatment in a short period of time on emergency department. The latter involves the evaluation of the cardiac vector velocity after the period of post-infarction electrical remodeling which may be useful in the follow-up of patients.
Conclusions: We have concluded that this method extends the concept of cardiac vector velocity and may be useful in the diagnosis of myocardial infarction.

1. Introduction

Early diagnosis of myocardial infarction (MI) is essential for implementing a rapid therapy to reduce possible complications or even fatal outcome. Currently, the gold standard for diagnosing a MI is a significant rise of plasma troponin levels between 12 and 24 h post myocardial injury [1]. However, this marker is elevated in a variety of other conditions and for this reason, this diagnostic criterion is supplemented with clinical judgment and several electrocardiographic (ECG) or vectorcardiographic (VCG) programs, mainly based on the presence of Q waves or changes in the ST segment [1–4]. These programs need to have high specificity to avoid unnecessarily treating patients who did not suffer a MI and high sensitivity, to quickly begin treatment of patients with early MI [5]. Unfortunately, although many modern algorithms (also including enzyme assays, body surface mapping and ECG with higher number of electrodes) might solve the first issue, it is difficult to increase sensitivity without loss of specificity [4,6,7].

Pathological alterations, provoked by fibrosis after myocardial injury, cause slow conduction by increasing axial resistance through effects on coupling between cells [8]. This fact may induce abnormal alterations in the dynamics of the cardiac vector and those alterations could be registered using the VCG which is very much utilized by physicians. Several studies have suggested the diagnostic usefulness of spatial velocity of the tip of the cardiac vector obtained during ventricular depolarization [2,9]. In the present work we have developed a novel algorithm to compute the angular and linear velocities during ventricular repolarization. We hypothesize that these parameters combined with the spatial velocity of the depolarization process can achieve high sensitivity and specificity simultaneously as discriminator of MI.

Abbreviations: AUC, area under the ROC curve; ICVV, index of cardiac vector velocity; MI, myocardial infarction; MI7, study group with MI whose VCG recording was made within the following 7 days after MI injury; MI45, study group with MI whose VCG recording was made after 45 days of injury.

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2. Materials and methods

2.1. Study population

We have used the Physikalisch-Technische Bundesanstalt (PTB) database which have been acquired at the Department of Cardiology of University Clinic Benjamin Franklin in Berlin, Germany, and has been provided to the users of PhysioNet. The data have been studied anonymously, using publicly available secondary data, therefore no ethics statement is required for this investigation [10,11]. The population consisted of 290 subjects including healthy volunteers and patients with different heart diseases. Subjects without clinical summary were excluded. Ages ranged from 17 to 87 years with a mean of 56; 81 of the subjects were female (28%). Three study groups were selected: Group 1, the control group, consisted of 52 subjects with no previous cardiovascular disease; "MI7" group, composed of 93 patients with MI with different infarct sizes and locations, whose VCG recording was made within the following 7 days after MI injury; and "MI45" group, composed of 46 patients with MI whose VCG recording was made after 45 days of MI. Each record includes 15 simultaneously measured signals: the conventional 12 leads together with the 3 Frank lead ECGs. Each signal was digitized at 1000 samples per second.

2.2. Theoretical background

This work is based on the study of the dynamics of the tip of cardiac vector, obtained from Frank leads *XYZ*. These leads define a three-dimensional space where we can describe the vector movement through two parameters: linear and angular velocities. The former quantifies the speed of linear displacement from one sample point to another one and the latter quantifies the rotation speed, i.e. the amount of time it takes to traverse an angle between two consecutive samples.

2.2.1. Linear velocity

The linear velocity in the QRS loop (so-called spatial velocity of depolarization) has been reported to be used as part of diagnostic criteria for inferior MI, but by itself it has low sensitivity [2,9]. The instantaneous values can be computed by differentiation

$$\vec{v}_i = (v_{xi}; v_{yi}; v_{zi}) = \frac{P_i(x, y, z) - P_{i+1}(x, y, z)}{Ts} \tag{1}$$

where P_i is the value of the tip of cardiac vector in *XYZ* space at *i*th sample point and *Ts* is the sampling period. Therefore, we obtain each $\vec{v}_i$ from Eq. (1) along any portion of the VCG signal path. Particularly in this work, we have evaluated not only the QRS loop (in order to obtain the classical spatial velocity), but also the linear velocity of the T-wave loop. We have proposed to characterize the resulting signal using two parameters:

– $v_{\max}$, quantifying the maximum linear speed:

$$v_{\max} = \max(|\vec{v}|) = \max(\|\vec{v}_i\|_2) \quad \forall i\text{th} \quad \textit{sample} \tag{2}$$

– $v_{E\alpha}$, quantifying the total energy of the signal through the 1-norm of each α axis ($\alpha = x$, y or z):

$$v_{E\alpha} = \|v_\alpha\|_1 = \sum_{\forall i} |v_{\alpha i}| \tag{3}$$

2.2.2. Angular velocity

Currently, in other research areas such as orbits or aerospace navigation, the rotation speed of a vector is not obtained through Euler matrices but rather through the quaternion algebra. Quaternions were first presented by Hamilton [12] over a century ago. It is well known that they play an important role in an alternative form for a rotation operator. Furthermore, quaternions are very efficient for analyzing situations which involve three dimensional rotations in terms of uncertainty propagation and data processing speed [13]. If we consider the tip of the cardiac vector as a point that rotates around the origin of the electrical axis of the heart, then we can apply the quaternion theory to compute the angular velocity.

For each pair of consecutive points P_i and P_{i+1} in *XYZ* space, normalized to the sphere of radius 1, we can define a quaternion associated with rotation angle θ:

$$\mathbf{q_i} = \cos\left(\frac{\theta}{2}\right) + \vec{u} \cdot \sin\left(\frac{\theta}{2}\right) \tag{4}$$

where the first term represents an amount of rotation and $\vec{u}$ is the rotational axis. In addition, considering the definitions of dot product and cross product

$$\begin{cases} P_i \cdot P_{i+1} = \|P_i\|_2 \cdot \|P_{i+1}\|_2 \cdot \cos(\angle(P_i, P_{i+1})) \\ P_i \times P_{i+1} = \|P_i\|_2 \cdot \|P_{i+1}\|_2 \cdot \sin(\angle(P_i, P_{i+1})) \cdot \vec{n} \end{cases} \tag{5}$$

we obtain

$$\mathbf{q_i} = (P_i \cdot P_{i+1}; P_i \times P_{i+1}) \tag{6}$$

which computes the quaternion for double-angle value.

Once each quaternion that describes rotation from P_i to P_{i+1} was obtained, the instantaneous angular velocity can be computed by solving the Poisson equation [14]:

$$\dot{\mathbf{q}}_\mathbf{i} = \frac{1}{2} . \vec{w}_i . \mathbf{q_i} \tag{7}$$

Finally, multiplying both sides by the inverse of the quaternion and bearing in mind that we obtained a double-angle expression, then

$$\vec{w}_i = \dot{\mathbf{q}}_\mathbf{i} . \mathbf{q_i}^{-1} \tag{8}$$

As in Section 2.2.1, we obtain several descriptors for the angular velocity along the T-wave loop on the VCG signal path:

– $w_{\max}$, quantifying the maximum angular speed:

$$w_{\max} = \max(|\vec{w}|) = \max(\|\vec{w}_i\|_2) \quad \forall i\text{th} \quad \textit{sample} \tag{9}$$

– $w_{E\alpha}$, quantifying the total energy of the signal through the 1-norm of each α axis ($\alpha = x$, y or z):

$$w_{E\alpha} = \|w_\alpha\|_1 = \sum_{\forall i} |w_{\alpha i}| \tag{10}$$

2.3. Algorithm

The *XYZ* signals were selected for each recording. A Butterworth high-pass filter (0.5 Hz, bidirectional) has been applied for baseline wander correction. We have defined two signal windows: The first in R-wave peak position ±60 ms and the second in T-wave peak position ±120 ms. In order to reduce high frequency noise in both signals, a Butterworth bidirectional low-pass filter (45 Hz and 20 Hz, respectively) has been used. QRS complexes and T-waves have been located using the wavelet-transform based method in [15].

Given that tissue damage constantly alters the conduction velocity in every beat [8] and that our hypothesis is based on the fact that these alterations may be measured from the VCG, we have calculated the velocity signals of Sections 2.2.1 and 2.2.2 on each beat and subsequently applied an average on 50 beats. For this purpose, it has performed an alignment of each signal based on minimizing the mean square error.

We have shown in Fig. 1 an example of $\vec{v}_i$ and $\vec{w}_i$ signals resulting from the algorithm in a scaled T-wave loop.