THE FIRST-ORDER HIGH-PASS FILTER INFLUENCES THE AUTOMATIC MEASUREMENTS OF THE ELECTROCARDIOGRAM

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ABSTRACT

We have studied the effects of the 0.05 Hz first-order high-pass filter (AC coupling) on the automatic measurements of the electrocardiogram (ECG). The standard 12-lead ECG of 1248 patients and the automated measurements of the AC- and DC-coupled (unfiltered) versions were compared. We found a strong, linear correlation between the QRS integral and the difference in ST-segment amplitudes, suggesting that the AC coupling alters the ST segment. The effect on the remaining part of the ECG was minimal. Medical professionals and developers of software for ECG interpretation should be aware of such high-pass filter effects, since they could be misinterpreted as pathophysiology or some pathophysiology could be masked by these effects.

Index Terms— AC coupling, Automatic measurements, ECG measurements, high-pass filter, ST segment

1. INTRODUCTION

Case reports have shown that high-pass filtering of an ECG-signal may mask a disease. García-Niebla et al. found that improper high-pass filtering led to a false-positive Brugada syndrome diagnosis in a patient [1]. Ruta et al. experienced the same in 2013[2]. Other scientists have examined the high-pass filter itself and its relation to the ECG waveform [3-7]. It is well known that the low-frequency content of the electrocardiogram (ECG) is of importance in relation to the interpretation of the ECG and especially the detection of acute myocardial infarction (AMI) [8].

The resting ECG may also be used for gatekeeper purposes, deciding which chest-pain patients go directly to the catheter lab and which go to the Emergency Room for further evaluation. Therefore, the quality of these ECGs directly influences the level of care for the patient, and the quality of every such recording must be optimal.

1.1. The AC coupling in Electrocardiographs

The AC coupling in electrocardiographs became necessary when digital recordings were introduced, due to a limited range for analog-to-digital converters (ADCs). The analog high-pass filter was introduced as a solution. In 1966, Berson and Pipberger did extensive research on the topic of first-order analog high-pass filters and their influence on the ECG waveforms. They concluded that high-pass filtering may influence the morphology of ST segment and T wave when the cut-off frequency of the filter is too high. Based on their research, they suggested a maximal cut-off frequency for a first-order high-pass filter of 0.05 Hz [9].

1.2. Tests and requirements for high-pass filters

Berson and Pipberger's suggestion was adopted in the 1967 AHA recommendations and kept in the 1975 version [10, 11]. In 1990, a special report from the AHA [12] suggested a rectangular pulse test to limit high-pass filters such that the distortions to the ST segment would be similar to those of a single-pole 0.05 Hz high-pass filter. The exact suggestions for a test were the following:

- "A 1 mV-sec impulse should not produce a displacement greater than 0.3 mV after the impulse."
- "For a 1 mV-sec impulse input, the slope of the response outside the region of the impulse should nowhere exceed 1 mV/sec."

This test was adjusted and accepted in the 1991 AAMI/ANSI standards for diagnostic ECG devices [13], so that the pulse now was a 0.3 mV·s pulse (3 mV·0.1 s). The offset requirement became 0.1 mV and the slope requirement became 0.3 mV/s. Both current American and international standards [13, 14] have the pulse test in exactly this form. The current specifications therefore directly date back to research made in 1966.

1.3. Distortions from the AC coupling

In 1966, it was clearly stated that the 0.05 Hz filter was a compromise that "the electrocardiographer has to live with", and that "ideally, records with a DC-response should be used". With the modern opportunities of the digital DC-coupled ECG recorder [15] and computer measurements, we wish to quantify the distortions to the ST segment caused by the AC-coupling with a 0.05 Hz first-order high-pass filter.

2. METHODS

2.1. Model

We propose a simplified model, by which the J amplitude is affected by the area of the QRS complex only. The time constant of the 0.05 Hz filter is 3.2 seconds, thus the QRS duration of usually less than 150 ms is not included in the model. The same argument can be applied for the QRS

TABLE I
AMPLITUDE DEVIATIONS IN RELATION TO QRS INTEGRAL

Measurement	Slope of best fit	R^2 for best fit	Offset for fit
P amplitude[a]	0.011 µV/(µV·s)	0.003 (p<0.001)	0.05 µV
Q amplitude	-0.025 µV/(µV·s)	0.000 (p=0.035)	0.24 µV
R amplitude	-0.039 µV/(µV·s)	0.002 (p<0.001)	-1.42 µV
S amplitude	-0.100 µV/(µV·s)	0.005 (p<0.001)	-0.28 µV
J amplitude	-0.277 µV/(µV·s)	0.649 (p<0.001)	0.82 µV
J10 amplitude	-0.276 µV/(µV·s)	0.804 (p<0.001)	0.80 µV
J20 amplitude	-0.271 µV/(µV·s)	0.807 (p<0.001)	0.89 µV
J40 amplitude	-0.261 µV/(µV·s)	0.796 (p<0.001)	0.95 µV
J60 amplitude	-0.245 µV/(µV·s)	0.756 (p<0.001)	0.98 µV
J80 amplitude	-0.232 µV/(µV·s)	0.706 (p<0.001)	0.96 µV
ST average	-0.253 µV/(µV·s)	0.825 (p<0.001)	0.89 µV
T amplitude[a]	-0.158 µV/(µV·s)	0.251 (p<0.001)	-1.68 µV

[a]Only for positive measurements. Negative amplitudes are measured separately and not shown.

morphology. The area of the QRS complex is called the QRS integral and is calculated with respect to the Q onset.

Our hypothesis is that a positive QRS integral creates an ST depression while a negative QRS integral creates an ST elevation and that there exists a linear dependency between the QRS integral and the offset produced.

The size of the J-deviation is heart-rate dependent, and based on simulations we expect a linear coefficient of $-0.277\ \frac{\mu V}{\mu V \cdot s}$ for a heart rate of 70 bpm and QRS duration of 100 ms. We further expect the 0.05 Hz filter to have an influence on the entire ST segment and the T wave as well, only the offset is smaller at a greater distance from the J-point. Therefore the linear coefficient will be smaller for the T amplitude.

2.2. Recovering DC signals

Berson and Pipberger suggested an inverting system for restoring the DC coupling [9]. This is just the interchange of the zero and the pole and can be done in the digital world as well.

1355 ECGs from patients with chest pain from the APACE study (Advantageous Predictors of Acute Coronary Syndromes Evaluation) [16] were used. An automatic lead quality assessment system [17] was used to determine which ten seconds of the ECG to use. For the filter to be relaxed, the snapshot would be taken at 20 seconds after all leads were connected if the global quality was better than 50 %. If the signal quality was below this level we would wait for the quality to improve since a physician would not start the recording either. This led to the exclusion of 16 signals (1 %) that did not meet these criteria. The number of initially included ECGs totals 1339.

Recordings were made using a commercial DC-coupled amplifier (CS-200 Excellence, Schiller AG, Baar, Switzerland) having a digital high-pass filter with a cut-off frequency below 0.05 Hz. The effects of this high-pass filter were reverted with the inverse filter to obtain DC-coupled electrocardiograms. A first-order digital high-pass filter with a cut-off frequency f_c=0.05 Hz was used to generate AC-coupled signals.

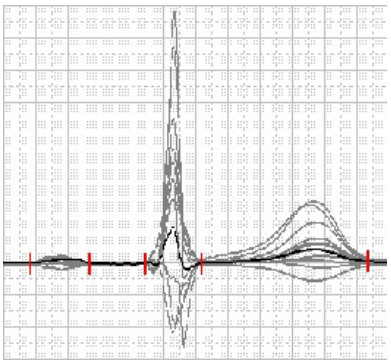

Fig. 1. Example of a butterfly 12-lead ECG with markers for P_{on}, P_{off}, QRS_{on}, QRS_{off} and T_{off} (left to right). The four segments may be denoted P wave, PR segment, QRS complex, and STT segment respectively. J_{40} amplitude is the amplitude 40 ms after QRS_{off} (J-point).

Today, most ECG amplifiers record less than 12 leads and construct the remaining channels digitally. This construction as well as the considered high-pass filtering are both linear operations. Thus, their order does not matter.

2.3. Comparing AC- and DC-coupled signals

Automatic measurement was performed using a commercial Resting ECG Measurement and Interpretation module (ETM, V1.12.0.0, Schiller AG, Baar, Switzerland). This module detects beats in the ECG using a QRS detector and classifies them based on their morphologies. An average beat is produced from the dominant class, and markers (P_{on}, P_{off}, QRS_{on}, QRS_{off}, and T_{off}) are determined via the use of vectorcardiography (VCG). Measurements are made on the averaged beat by the use of these markers and other VCG information such as spatial velocity.

A full analysis of each ECG was run, and the differences between the analyses of AC- and DC-coupled signals were analyzed through the use of histograms, mean values, and standard deviations of the differences. The correlations between the QRS integral and the differences in P, Q, R, S, J, J10, J20, J40, J60, J80 and T amplitudes and ST average were also calculated (reference QRS_{on}). ST average is the mean value from the J-point to the midpoint between J and T_{max}. Jx amplitude is the amplitude x milliseconds after the J-point. Since the J amplitude was of primary interest, a recording was excluded if the difference in measurements of the QRS-duration exceeded 5 ms (91 cases, 6.8 %). The J-point is the end of the QRS complex (junction with ST segment, Fig 1) and if the J-point is determined to be located only a few milliseconds too early, large measurement differences may occur because of the steep slope of the QRS complex. These differences are not caused by the filtering, but rather by the measuring technique.

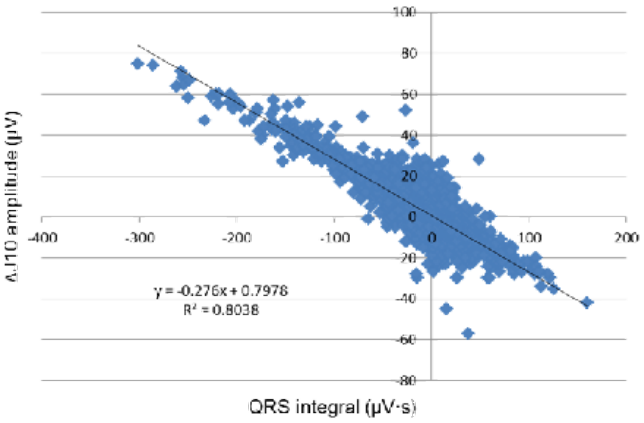

Fig. 2. Scatter plot of differences in J10 amplitude against QRS integral. A correlation is clear and statistically significant ($p<0.001$). The squared correlation coefficient is $R^2=0.804$, which is better than for the J-amplitude, since the point is located further away from the steep QRS-complex.

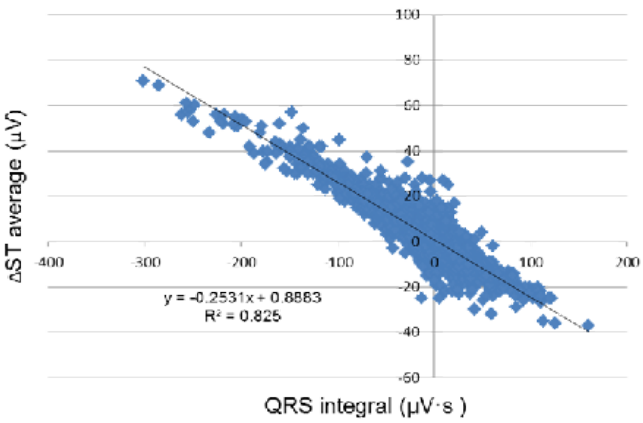

Fig. 3. Scatter plot of differences in ST average against QRS integral. A correlation is clear and statistically significant ($p<0.001$). The squared correlation coefficient is $R^2=0.825$. The ST-average is very stable and resistant to noise.

The fail criterion was defined as a deviation greater than ± 25 μV between AC and DC ECGs. This choice was made based on limits in the current American and International Standards [13, 14].

3. RESULTS

Fig. 2 shows the link between QRS integral and J10 amplitude. The J10 amplitude is more stable than the J amplitude because it is located further away from the QRS complex where large deflections are found.

The more stable ST average also shows a linear dependency on the QRS integral (Fig. 3). The linear coefficient is $-0.2531\ \frac{\mu V}{\mu V \cdot s}$ which places the ST average between J40 and J60. This would suggest a reasonable average ST-segment duration of 100 ms.

In Table I, values for best fit first order lines for the scatter plots of QRS integral against differences in measurements (as in Fig. 2) are shown along with their R^2 values. According to the model, the offset should be minimal and has been included for validation purposes only.

Fig. 4 shows the expected ST-segment deviations for a constant heart rate of 71 bpm using rectangular QRS complexes of 100 ms duration and 1 μV·s area. For the 1248 ECGs, the mean QRS-duration was 94.6 ms and the mean heart rate was 71.1 bpm. Also shown are the ST deviations that were found experimentally.

4. DISCUSSION

4.1 Magnitude of errors

The error limit was defined as ± 25 μV which was reached for a substantial amount of leads (1.0 %). From all 1248 included ECGs, 49 ECGs (3.9 %) had at least one lead with a J amplitude above this limit while having a QRS integral that was greater than ±90.25 μV·s in that lead. For J10 amplitude, the number is 50 (4.0 %) and for ST average the number is 45 (3.5 %). Further, some signals display deviations of up to three times this limit in concordance with the model.

Since the offset is heavily dependent on the QRS integral, some patient groups are more affected by the 0.05 Hz filter than others. Patients with a Left Bundle Branch Block (LBBB) have ECGs with large QRS integrals in some leads. The diagnosis of myocardial infarction (MI) in the presence of LBBB is already recognized as a special case [18], and this may in part be accounted for by the 0.05 Hz filter. Patients with Left Ventricular Hypertrophy (LVH) typically have ECGs with high amplitudes, which also lead to larger QRS integrals.

4.2. Slope of the ST segment

Filters are required not to alter the slope of the ST segment too heavily. A filter may be limited by the production of a slope of less than 300 μV/s for the 0.3 mV·s pulse. The estimated ST-slope shown in Fig. 4 is $0.41\ \frac{\mu V}{\mu V \cdot s \cdot s}$ which would produce a slope of 123 μV/s for the pulse in question. Of the 1248 ECGs, not a single signal had an absolute QRS integral above 300 μV·s, meaning the test is robust as a worst-case scenario test. In the clinical setting, the exact value of the slope of the ST segment is not important; rather it is sufficient to know whether the ST segment is descending, ascending, or horizontal. For a paper resolution of 10 mm/mV and 25 mm/s, a "worst-case" slope of 123 μV/s corresponds to 2.8 degrees. We do not believe that this effect would change the classification of a significant number of ST segments; however, further studies are

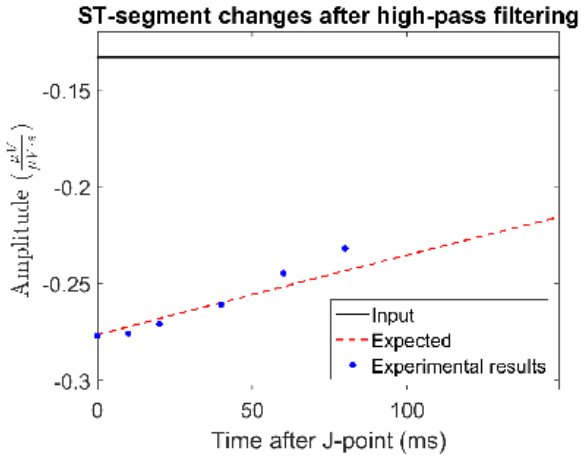

Fig. 4. Expected and actual results from using a 0.05 Hz first-order high-pass filter. The reference point for the horizontal axis is the J-point. The reference point for the vertical axis is the QRS onset for the filtered signal. The input pulse was normalized to 1 µV·s and has a duration of 100 ms. The exact coordinates for each experimental result can be read from Table I.

required to say anything with certainty.

4.3. Model validation

A linear dependency of the amplitude offset after the QRS complex to the area of the QRS complex has been found as predicted by the model. This linear dependency is furthermore predicted to be heart rate dependent, which explains a large portion of the deviations from the average trend. Mean heart rate was found to be 71.1 bpm (from the 1248 signals) with a standard deviation of 15.3 bpm. Dividing the signals into subclasses based on heart rates requires a larger database than was available. Future work should cover this aspect.

As listed in Table I, the linear coefficient for the J amplitude was found to be -0.277 $\frac{\mu V}{\mu V \cdot s}$, which is what we expect for a 70 bpm ECG. With a mean heart rate of 71.1 bpm, the trend appears to be just as it was modeled to be.

Differences in measurements due to unequal determination of markers from AC-coupled to DC-coupled signals occur. These differences are largest for the J amplitude. When significant power line interference (50 Hz) is present, this effect becomes even more important for point measurements such as the J and J10 amplitudes, since only 5 milliseconds are a quarter of a cycle. The ST average is much more resistant to such noise.

Further, all absolute offsets for best fit curves are below 2 µV, in accordance with the model.

4.4. Model limitations

For the ideal signal with constant heart rate, easily determined markers, and minimal (power line) noise, the model will still exhibit limitations in its precision. The QRS morphology plays a limited role of only a few percent of the total ST-deviation.

The model does not account for varying heart rates, nor does it incorporate differences in QRS morphology or ventricular extrasystoles. However, for these cases, the ST-segment suppression or elevation may not be of particular interest. The model does not acknowledge that the P and T waves may influence the offset.

4.5. The 300 µV·s pulse test

The 300 µV·s test was designed to limit the effects of a high-pass filter to that of the 0.05 Hz first-order high-pass filter. As we have shown, this filter may distort the ECG waveform, raising reasonable doubts as to the effectiveness of the test, at least with the current limits. Alternative tests, such as the tests with calibration and analytical artificial ECGs suggested by the IEC [14], may provide a more reliable test.

4.6. Impact on the ECG

Beside the J and ST amplitudes, also the T wave was affected by the filtering. However, while these magnitudes of the changes in a clinical sense constitute critical differences in the ST segment, such magnitudes of changes are less important in the T wave. Further, the amplitude of the changes to the T wave is smaller than of those to the ST segment.

A similar effect was not found for the P, Q, R, and S amplitudes, and any changes that did occur had no correlation to the QRS integral or QRST integral.

5. CONCLUSION

The 0.05 Hz first-order high-pass filter has been reviewed. Application of the filter in some cases changes the amplitude of the J-point by more than 25 µV.

For resting ECGs, a linear correlation was established between the QRS integral and the change in J amplitude. This effect persists throughout the ST-T segment with decreasing amplitude.

It is important to know the effect that the 0.05 Hz filter has on the ECG, particularly in case of large positive or negative QRS areas, when interpreting the ECGs.

We conclude that significant differences may exist between DC-coupled and AC-coupled recordings. In the ST-segment the problem is the amplitude alterations while the slope of the segment is not an issue. The DC-coupled ECG is to be preferred, since such alterations are not physiological. Medical professionals, who have previously interpreted AC-coupled ECG signals, should be made aware of the alterations that are made to the ST segment, as should manufacturers of ECG interpretation programs.

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