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A review of electrocardiogram filtering

Shen Luo, PhD,* Paul Johnston
Cardiac Science Corporation, Deerfield, Wisconsin, USA
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Abstract Analog filtering and digital signal processing algorithms in the preprocessing modules of an electrocardiographic device play a pivotal role in providing high-quality electrocardiogram (ECG) signals for analysis, interpretation, and presentation (display, printout, and storage). In this article, issues relating to inaccuracy of ECG preprocessing filters are investigated in the context of facilitating efficient ECG interpretation and diagnosis. The discussion covers 4 specific ECG preprocessing applications: anti-aliasing and upper-frequency cutoff, baseline wander suppression and lower-frequency cutoff, line frequency rejection, and muscle artifact reduction. Issues discussed include linear phase, aliasing, distortion, ringing, and attenuation of desired ECG signals. Due to the overlapping power spectrum of signal and noise in acquired ECG data, frequency selective filters must seek a delicate balance between noise removal and deformation of the desired signal. Most importantly, the filtering output should not adversely impact subsequent diagnosis and interpretation. Based on these discussions, several suggestions are made to improve and update existing ECG data preprocessing standards and guidelines.

Keywords: ECG filter; Frequency domain; Aliasing; Linear phase; Magnitude distortion; Ringing artifact; ECG standards

Introduction

Electrocardiographic devices, including interpretative electrocardiogram (ECG) units, stress ECG systems, bedside ECG monitors, ambulatory recorders, and others, are primary tools used by routine clinical practices for conducting cardiovascular diagnostic and monitoring procedures.[1] The accurate recording and precise analysis of the ECG signals are crucial due to their extensive applicability and also the high-performance expectations of medical professionals. Accurate interpretations of ECGs have always relied heavily on state-of-the-art signal processing.

The preprocessing modules for an electrocardiographic device contain multiple levels of signal manipulation and detection routines, which start by converting analog signals into digital data that is used for analysis, interpretation, and presentation (display, printout, and storage). This article aims to discuss a key ECG processing topic: ECG filtering and data accuracy in a modern electrocardiographic device.

There are 4 typical filter processes in an ECG device: (*a*) anti-aliasing and upper-frequency cutoff, (*b*) baseline wander suppression and lower-frequency cutoff, (*c*) line-frequency rejection, and (*d*) muscle artifact reduction. Use of additional proprietary filter algorithms (such as cubic spline technique,[2] time-varying muscle artifact filter,[3] source consistency filter,[4] etc) to manipulate waveform data is not included in this discussion. In this review paper, we first introduce the basis of filtering in the time and frequency domains because these concepts are fundamental to the discussion of data accuracy. For each filter process, we then pick 1 or more important topics regarding inaccuracy issues to summarize a compliance review and a discussion of the implications.

Reviewing the basis of ECG filtering

Filtering, magnitude (or amplitude), and phase distortions

A given signal in the time domain can mathematically be represented in terms of its magnitude and phase responses in the frequency domain, whereas a given filter with its impulse response in the time domain can be characterized by its magnitude and phase responses in the frequency domain (see Fig. 1).

Except for some very special situations (such as an all-pass filter), a filter is generally designed to attenuate or remove some frequencies from the input data. We expect that a filter only removes the noise without changing the desired signal. In the real world, noise and desired signals often

* Corresponding author. Cardiac Science Corporation, 500 Burdick Parkway, Deerfield, WI 53531, USA.
E-mail address: sluo@cardiacscience.com

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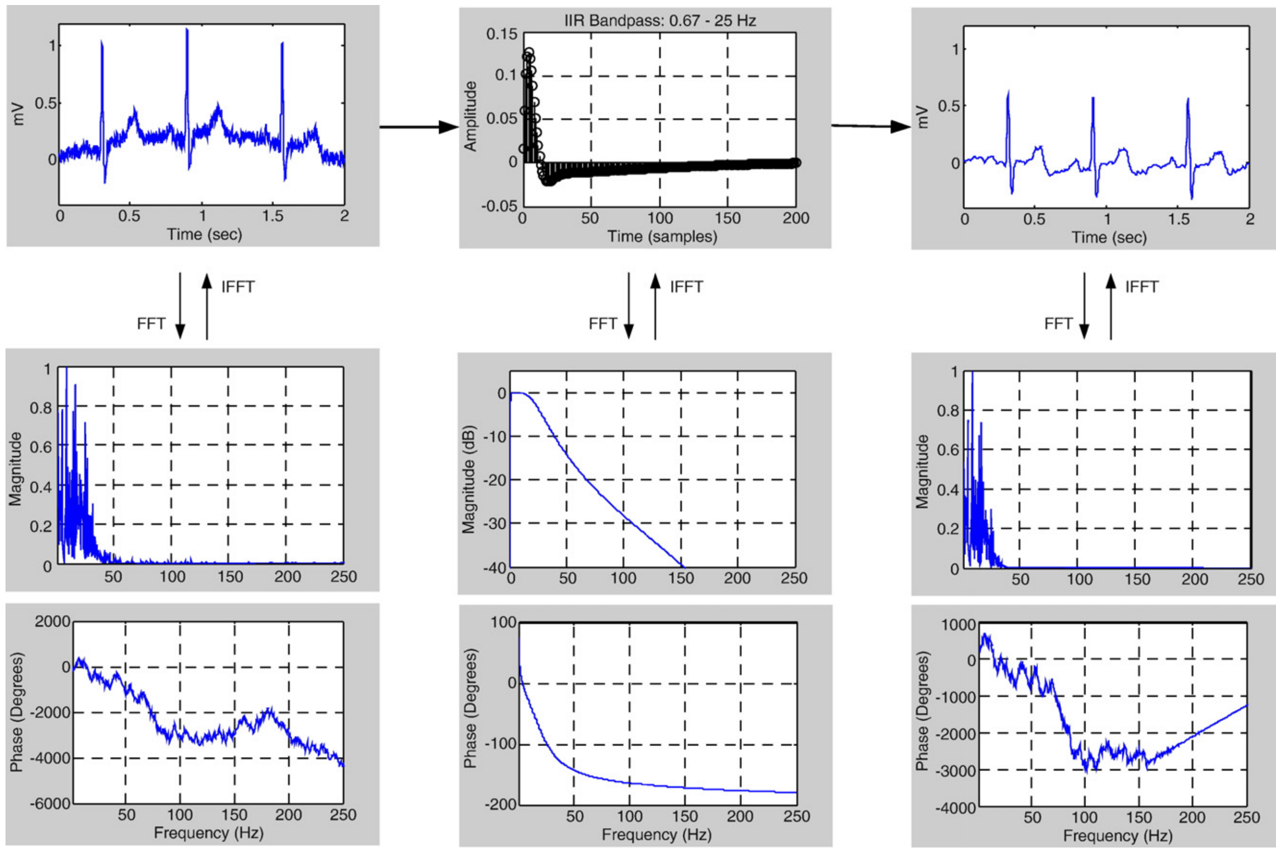

Fig. 1. Both signals and filters can be mathematically converted (mapped) from the time domain to frequency domain, or vice versa. When an ECG signal (left) goes through a filter (middle), the filtered ECG trace (right) could be altered, although noise is reduced. The frequency response, including both magnitude and phase responses, can help us to evaluate and comprehensively design a better filtering system to minimize the distortions and maximize the signal to noise ratio.

overlap in the frequency domain. As a result, when a filter attenuates the frequency components of noise, the overlapping desired signal can also be impacted, causing magnitude distortion of the desired signal.

In addition to the magnitude response, the phase response is another property of a filter. Phase distortion occurs when a filter's phase response is not a linear function of frequency so that the phase shift is not directly proportional to the frequency. Phase distortion introduced by a filter could produce a significant impact on data accuracy (examples of phase distortion can be found later in Figs. 3 and 4). As a simple expression, a digital infinite impulse response (IIR) filter has nonlinear-phase response, whereas a digital finite impulse response (FIR) filter can be designed to have a linear-phase characteristic over the frequency range of interest.[5] Regarding other more general comparisons between IIR and FIR filters, please refer to signal processing and filter design textbooks (such as those of Oppenheim and Schafer[5] and Jackson[6]). Also, we will discuss some strengths and weaknesses, especially for the FIR filter, in the individual ECG filtering process sections that follow.

Data sampling, Nyquist frequency, and anti-aliasing

A modern ECG device is basically a digital system. After preliminary processing by the front-end module, the analog ECG signal is immediately converted into a digital form (Analog to Digital Conversion [A/D]) at a particular sampling rate or frequency for further usage.

The Nyquist frequency, also named the folding frequency, is half the sampling rate. Per the sampling theorem, the bandwidth of the input signal should not be greater than the Nyquist frequency. Signal frequencies higher than the Nyquist frequency will encounter a "folding" about the Nyquist frequency and map false components back into lower frequencies. An example is a frequency component at 10 Hz above the Nyquist frequency, which is folded backward to 10 Hz below the Nyquist frequency. This effect is called aliasing.

The Nyquist frequency is a key concept in the initial data sampling (with A/D conversion) and resampling processing (such as further down-sampling).[5] To prevent aliasing interference, the signal must be band-limited. The anti-aliasing process is to use a low-pass filter (LPF) to reject the unwanted frequencies (equal to and greater than the particular Nyquist frequency) of the input signal before sampling or resampling.

Oversampling and down-sampling

Oversampling technique is used in many ECG devices. Oversampling simply refers to an initial A/D conversion sampling rate f_{os}, which is many times higher (eg, 8000 Hz) than the final data resolution target sampling rate f_s (eg, 500

Hz) that is used for further processing of the ECG signal. Note that usually $f_{os} \gg f_s$. The oversampled data with higher time resolution could be used for applications like pacemaker detection.[7]

From the oversampling rate f_{os} to the target sampling rate f_s, there is a decimation process, including filtering by an LPF and then resampling (down-sampling) the resulting smoothed signal at a lower sampling rate.[5,6] As mentioned in "Data sampling, Nyquist frequency, and anti-aliasing" section, use of an LPF avoids aliasing during data stream rate changes.

Oversampling has several other advantages such as facilitating pacemaker spike presentation on ECG printouts (with reduced amplitude from an LPF, see Appendix I), quantization error improvement if needed regarding the precision of the least significant bit,[8] and lower order analog anti-aliasing filter implementation (see Appendix II).

Cutoff frequency and 3-dB attenuation

We often simply use cutoff frequency f_c to describe a filter. For example, a 150-Hz LPF means that this filter passes low-frequency signals and has a cutoff frequency at 150 Hz. The term cutoff comes from an ideal filter that can be realized theoretically through a rectangular function in the frequency domain, where there is not a transition and the cutoff frequency is the function's edge or boundary between a passband (signal is unchanged) and a stopband (signal is completely eliminated).

Unlike an ideal filter, a real filter has a transition band from the passband to the stopband. Using the same expression, cutoff frequency f_c is defined as the frequency at which a 3-dB attenuation or 30% reduction occurs (but not completely cutoff) from the passband for magnitude response control. Therefore, a 150-Hz filter or a filter with 150-Hz cutoff frequency has a 3-dB "cutoff" or actual attenuation at 150 Hz. In broad terms, the band from the cutoff frequency where the magnitude response turns the corner to a frequency (at or before the Nyquist frequency) where the response reaches its bottom maximum attenuation can be viewed as the transition. With viable roll-off specifications applied in a real filter system, the Nyquist frequency could be significantly greater than the upper-frequency cutoff. Because 150 Hz is the recommended cutoff,[1,9] 250 Hz should be a common choice for the Nyquist frequency and that sets the digital data stream sampling rate at 500 Hz.

f_c is important in the frequency domain, but often, more specifications should be considered so that a panoramic picture of a filter process is taken into account by reviewing items such as the following:

(*a*) Flatness of the passband (such as over a range of 1-40 Hz for ECGs)
(*b*) The shape of the transition band
(*c*) The stopband
(*d*) Phase response

Inherent inaccuracy issues

Inherent noise and inaccuracy issues that can result from filter designs, such as aliasing, distortion, and ringing, can exist in devices and negatively impact interpretation of the ECG. Some problems can result from limits or side effects of the filter itself, whereby other degrading sources might be due to a mixture of filter design, implementation, and possible issues encountered in comprehensively testing per standards and guidelines.

Fidelity and the filter design goal

Filters are specified in the frequency domain, but the time domain response is what is seen in the signals. Although the result of filtering is judged by the fidelity with which it represents the original ECG signal,[9] the practical design goal, which is slightly different from a straightforward criterion in pursuit of signal fidelity, is often to try to find a good tradeoff between less magnitude distortion and noise reduction, or a better signal to noise ratio, while still minimizing or avoiding phase distortion.

LPF for aliasing and upper-frequency cutoff

Background

There are often 2 processes for anti-aliasing filtering: an analog LPF located before A/D conversion and a digital decimation LPF after digitization. The analog LPF is for the initial oversampling process, whereas the digital decimation LPF is for the down-sampling process. If either of them has weak attenuation at their individual stop bands, aliasing could appear. One practical scenario is that an ECG unit and another medical device are hooked up to a patient at the same time. The second device outputs an active signal with frequencies slightly higher than the target Nyquist frequency of the first ECG unit and couples apparent noise to the ECG input. Aliasing artifacts can be observed from the ECG printouts when a weak digital anti-aliasing filter is used.

The analog LPF has 3 other functions regarding frequency response: provide a flat passband, setup oversampling cutoff f_{oc} greater than 150 Hz (normally it is set much higher than 150 Hz due to oversampling) and minimize nonlinear-phase response. Similarly, a decimation filter provides a flat target passband and controls upper-frequency cutoff f_c = 150 Hz with 3-dB (30%) attenuation and, if possible, offers linear-phase response in the passband and even the transition band.

Compliance review

Currently, there is no specific reference to guide and check the performance of the analog LPF. Also, standards do not provide clear requirements to check for aliasing problems from the decimation filter. For diagnostic devices, American National Standard Association for the Advancement of Medical Instrumentation (AAMI) EC11[10] and International Standard IEC 60601-2-51[11] request that the output amplitude remains within a range between +10% and −100% of true amplitude when varying the input signal frequency from 150 to 500 Hz. The definition of this requirement could cause confusion.

It is well known that all frequencies for digital processing are relative to the sampling rate. For example, if the sampling frequency f_s = 1000 Hz, a filter's transition roll-off can be developed in a straightforward way so that it would offer close to 100% suppression at a frequency that can be located anywhere between 150 Hz (realistically >150 Hz) and 500 Hz. This anti-aliasing system design will perform well because the Nyquist frequency $f_s/2$ = 500 Hz. However, with a typical interpretive device that uses a data stream at the target sampling frequency f_s = 500 Hz, the problem is that the standards do not define output amplitude reduction from the $f_s/2$ point (250 Hz) up to 500 Hz where aliasing could occur in any common design.

We also noticed that Test C of Table 114 of IEC 60601-2-51 states "+10%/−50%" when input is from 100 to 150 Hz. In connection with Tests B and D, this requires that the upper-frequency cutoff of an interpretive device be up to 100 Hz rather than 150 Hz. We treated it as a key typographical mistake when comparing to the guidelines[1,9] as well as Table 4 in EC11 that correctly states "+10%, −30%."

Discussion of implications

To reject aliasing noise, an analog LPF should reject frequencies at and greater than $f_s/2$ if the target sampling frequency f_s is used in the A/D conversion. More often, an oversampling technique does apply to an ECG device. For this situation, an analog LPF should eliminate frequencies near the oversampling frequency f_{os} (not at the oversampling Nyquist frequency $f_{os}/2$; see Appendix II) while providing a flat passband greater than the target f_c. Also, it should have a near-linear phase for the passband.

If there is only 1 stage from f_{os} to f_s, the decimation filter should reject any frequencies at and greater than $f_s/2$, rather than achieving −100% at the fixed 500 Hz without a specific f_s. If f_s = 500 Hz, for example, the decimation filter should be designed with linear phase, small ripple in the passband, upper-frequency cutoff f_c at 150 Hz, and strong attenuation at $f_s/2$ (a response like filter A, not B, in Fig. 2).

There may exist more than 1 decimation (down-sampling) stage from f_{os} to f_s. For instance, we wish to decimate by a factor of 16. Rather than decimate immediately by M = 16, we will decimate by a factor of M_1 = 4, then by a factor of M_2 = 4. The advantage is that the computational cost would be reduced significantly.[12] Also, for ECG applications, the data stream after the first decimation (eg, 2000-Hz data if the oversampling is 8000 Hz) can be used for some special means, such as high-frequency ECG analysis or other research purposes. For this scenario, therefore, these filters should reject frequencies at and greater than their individual Nyquist frequencies in every instance, and the final frequency sweep in the system level should have a similar performance as filter A of Fig. 2.

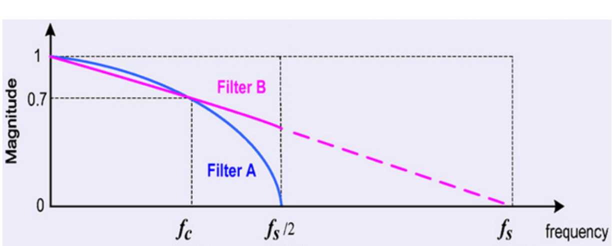

Fig. 2. Magnitude response of an ECG decimation LPF when the sampling rate is 500 Hz. Filter A has a strong attenuation at and greater than 250 Hz, whereas filter B has very weak attenuation at the Nyquist frequency.

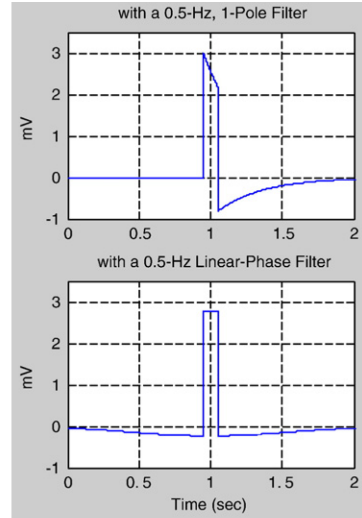

Fig. 3. Input is a single 0.3-mV·s rectangular impulse. Compared with the output of a 0.5-Hz 1-pole filter with nonlinear phase (top), the output of a 0.5-Hz filter with linear phase (bottom) has an offset amplitude (mimics the STj level) below the 0 level, but it is the same amplitude compared with the pulse onset (mimics the PR segment's isoelectric level). The dips around the output pulse ensure that the average value of the output within the window is zero.

Finally, the upper-frequency cutoff in IEC 60601-2-51 should be changed from 100 to 150 Hz, which is important especially for pediatric ECGs. In addition, if the upper-frequency cutoff is notably greater than 150 Hz when the sampling rate is 500 Hz, it can cause larger overshoot and ringing of the step response in the time domain due to the sharp attenuation in the frequency domain when rejecting frequencies toward a 250-Hz target. Narrow QRS complexes may also contain overshoot and ringing from this effect.

High-pass filter for baseline wander suppression and lower-frequency cutoff

Background

There are 2 important milestones for the technical requirements of lower-frequency cutoff in ECG instruments. The first is the 1975 American Heart Association (AHA) recommendations.[13] Based on an analog single-pole filter with insignificant distortion of the ST segment and QT interval,[14] the committee recommended a lower-frequency cutoff (3-dB down) of 0.05 Hz. Although this cutoff is too low to suppress most baseline wandering, it is still a classical reference basis to compare the performance of a digital filter at the current time.

The second is the 1990 AHA recommendations by an Ad Hoc Writing Group Committee of the AHA.[9] Based on availability of digital filters and electrocardiography systems, considering that the longest RR interval corresponds to the lowest frequency components of the ECG, this document recommended a lower-frequency cutoff (3-dB down) at 0.67 Hz that corresponds to a heart rate of 40 beats per minute (bpm), and a 1-mV·s testing impulse for displacement and slope evaluation. Also, it requires less than 0.5-dB ripple over the range of 1 to 30 Hz.

Here we briefly review available high-pass filters (HPFs) that can be used to suppress baseline wander and setup lower-frequency cutoff. For an analog HPF, distortions of the ST segment will increase as the cutoff frequency increases above 0.05 Hz. There is a similar issue for a digital filter with nonlinear phase. Like its analog counterparts, an IIR digital filter with 0.5-Hz cutoff, for example, can make marked distortions to ST segments, although it has straightforward implementation especially when using design tools. On the other hand, a FIR digital filter can be designed with a linear phase, and the ST-segment distortion due to the phase issue is then completely gone. This filter is better able to preserve the fidelity of the ST-segment levels even when the cutoff is 0.5 Hz or higher. A linear FIR filter has longer delay and needs special design considerations.[9,15] These 3 filter designs can work in real-time situations. A fourth filter design uses 2 IIR filters with identical designs, filtering once in a forward direction and once in the reverse direction.[9,16,17] Using signal processing terms, it is a zero-phase filter, a special case of the linear-phase filter. Therefore, it does not have a phase distortion problem. Because this application requires the entire data segment to be available for reverse filtering, it cannot be accomplished in real time.

Compliance review

With some modification and harmonization, the basic concepts of the 1990 recommendations were adopted into the current American Standard AAMI EC11 and international IEC 60601-2-51, both for diagnostic ECG units, as well as AAMI EC13[18] and IEC 60601-2-27,[19] both for cardiac monitoring devices.

One key examination from these harmonized standards is to check the displacement level (ie, the ST-segment deviation) surrounding a single 0.3-mV·s impulse (3 mV for 100 milliseconds): this impulse shall not produce a displacement greater than 0.1 mV. Then the question becomes: What should be the reference for the displacement limit measurement—the 0 level or the level of the impulse onset (to mimic the PR segment)? Referencing Fig. 3, for a filter with a nonlinear phase, these levels are the same, but they are different for a filter with linear phase.

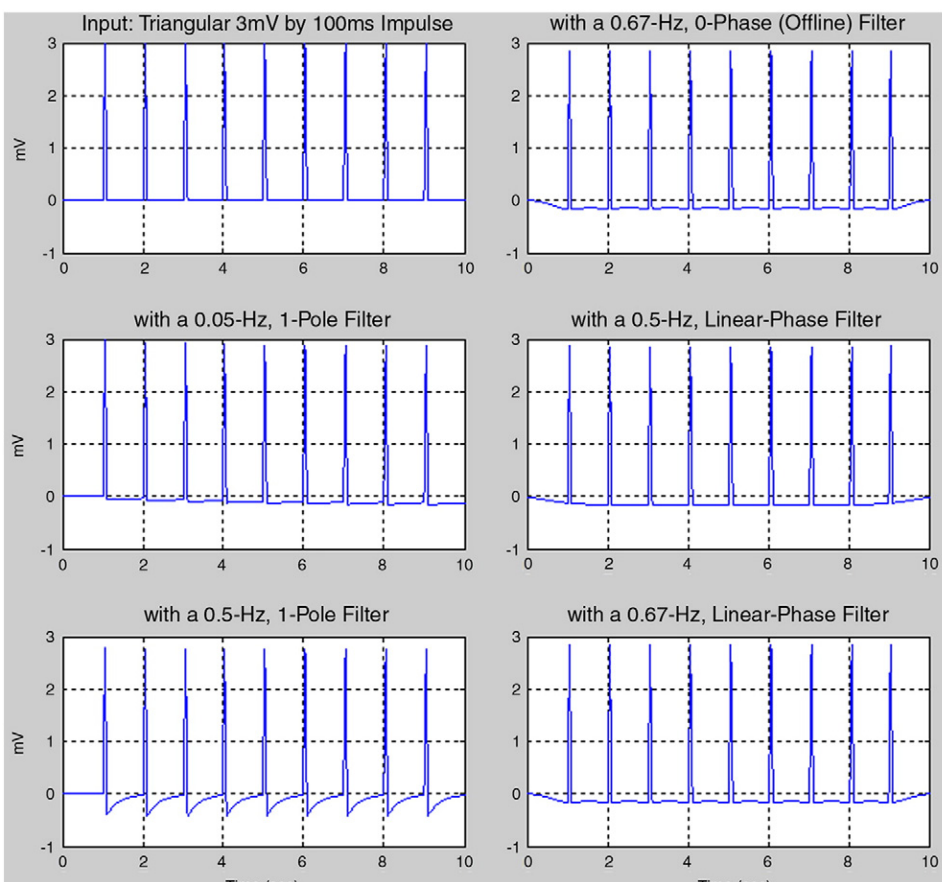

Fig. 4. Input (top left) and 5-HPF filter outputs at 60 cycles per minute (cpm) rate. Middle left: a 0.05-Hz single-pole IIR filter output has an increasing displacement during the 10 seconds below the 0 level (160 μV at 60 cpm by the end of 10 seconds). If the rate is 120 cpm, the displacement will be 310 μV below the 0 level. Bottom left: a 0.5-Hz single-pole IIR filter output introduces severe distortion due to the nonlinear phase, including an "overshoot" and a quick rise after the pulse offset. Top right: a 0.67-Hz forward-backward filter output (a zero-phase offline filter) has the same performance as the 0.67-Hz linear-phase filter bottom right. Middle right and bottom right: outputs from the linear-phase filters (0.5 and 0.67 Hz, proprietary) have nice flat baselines.

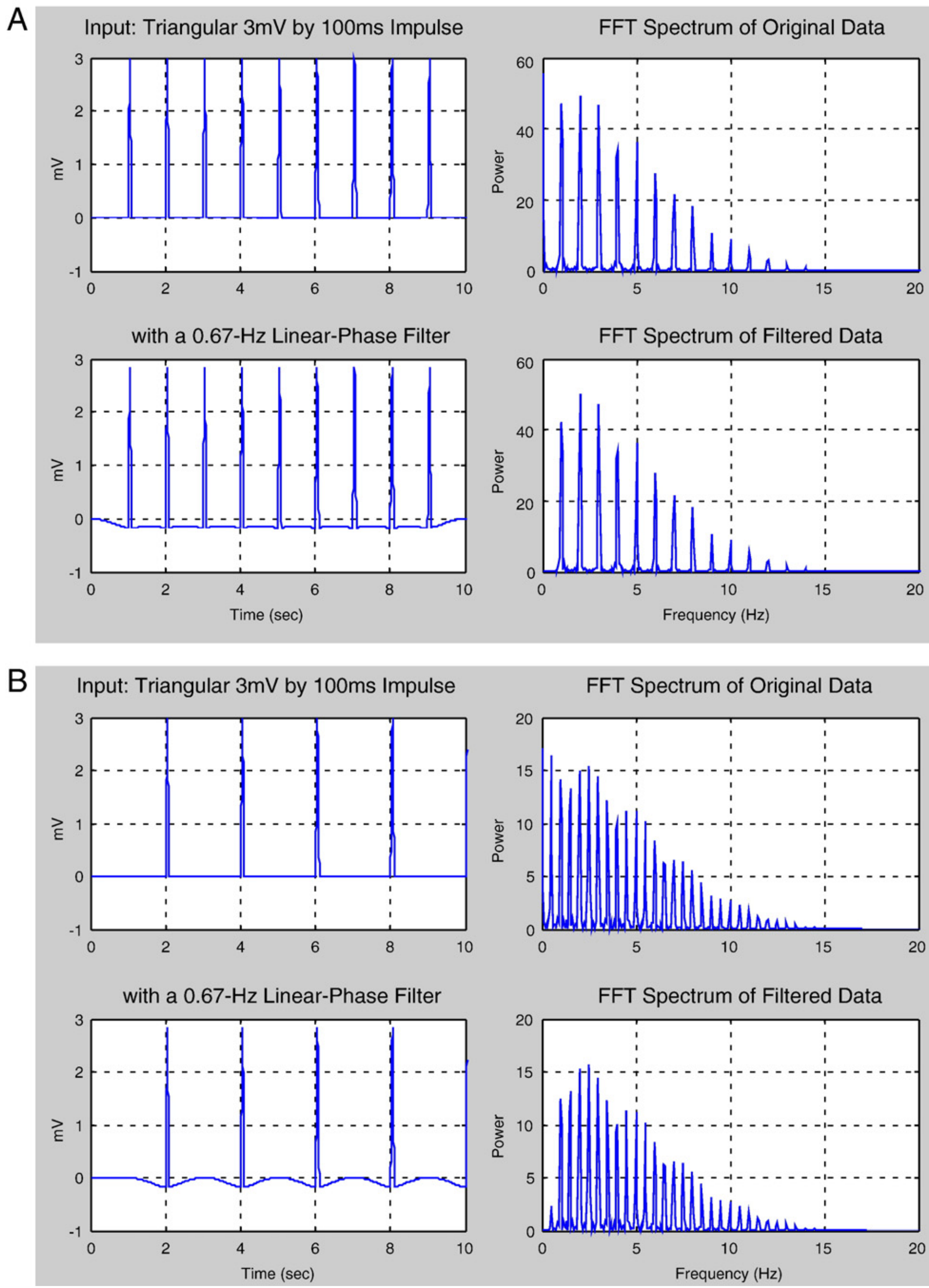

Fig. 5. Time and frequency domain changes of a 0.67-Hz linear-phase filter. At 60 cpm (A), the filter removes the DC, which results in the pulses plunging some of their portions into the negative voltage zone and forms a new steady-state level (left column). Also, frequency analysis (right column) shows the filter attenuates a small amount of the components around 1 Hz, which makes the filtered output nearly flat lines between the triangular waves. At 30 cpm (B), output has "dips" between 2 output triangular waves (left column). The slower the rate is, the deeper the dips are, and the baseline segments have more slope change. Frequency analysis (right column) shows the filter eliminates most components at 0.5 Hz, which causes the curving segments between 2 triangular waves. It is a magnitude distortion of the filter.

Therefore, the question is whether the reference point from this analog filter test needs to be updated for linear-phase digital filter testing.

Furthermore, the standards require −3 dB at 0.67 Hz for monitoring ECG devices (Table 3 of AAMI EC13 and Clause 50.102.8 of IEC 60601-2-27) and −0.9 dB (10%) at 0.67 Hz for interpretive ECG units (Table 4 of AAMI EC11 and Table 114 of IEC 60601-2-51), in addition to the displacement requirement of 0.1 mV or less for a single 0.3-mV·s impulse (as above). If using the 0 level as the reference, the maximal cutoff frequency for a single-pole filter to meet this requirement is 0.05 Hz[17]; for a linear-phase filter, it is around 0.25 Hz (far from 0.67 Hz) according to our pulse simulation. However, if the impulse onset as the reference is used, the results will be quite different, especially for a filter with linear phase.

Some simulation results

To study HPFs, we chose consecutive triangular impulses (0.3 mV·s) for imitating ECG beats at a repetition rate rather than a single-impulse test (see Figs. 4, 5, 6, and 7).

Discussion of implications

Choosing consecutive triangular impulses at a repetition rate to imitate ECG beats could be a better approach in examining the behaviors of nonlinear and linear-phase filters as well as evaluating the clinical implications.

Linear phase is the key for a baseline wander control HPF to minimize ST-segment distortion (Fig. 4). An HPF with nonlinear phase at higher cutoff is not suitable for clinical applications. By contrast, an HPF with linear phase or zero-phase response at higher cutoff can prevent "overshoot" after the QRS and distortion of the ST-segment slope because the phase shift is directly proportional to frequency.

Higher upper-frequency cutoff of a linear-phase filter could cause magnitude distortion when the heart rate is very low (marked bradycardia or pause) as shown in Fig. 5. It slopes the T-P segments for biologic ECGs (see simulation results in Fig. 6). It has an impact when using the T-P segment as the reference isoelectric level in measurements.

In addition to interpretive devices, consider using the limit of −0.9 dB, not −3 dB at 0.67 Hz, for a cardiac monitor device producing ST-segment measurements. A 0.5-Hz linear-phase filter with a correct design can meet the goal of 0.9-dB attenuation, and this frequency cutoff should be a better tradeoff.

On the other hand, a linear-phase filter with 0.67-Hz cutoff frequency may also be acceptable for stress ECGs. However, the attenuation at 1.0 Hz should be controlled to no more than −0.5 dB (5% attenuation) to reduce magnitude distortion as shown in Fig. 7.

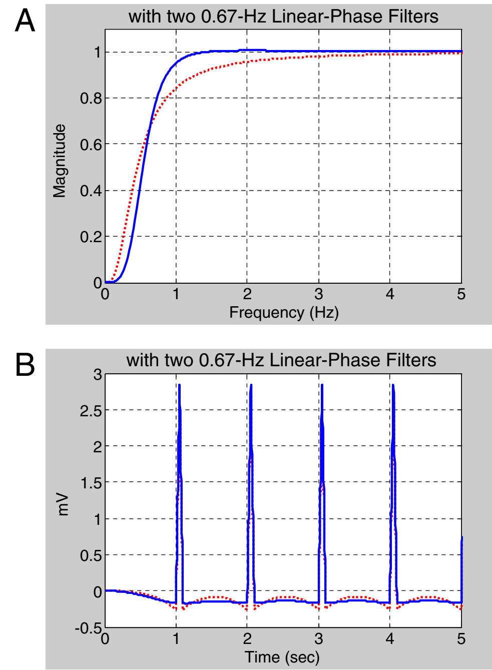

Fig. 7. Magnitude distortions from linear-phase filter design. Both linear-phase filters have cutoff frequencies at 0.67 Hz (A), but the performance in the time domain is quite different (B). From the magnitude responses, the corner behaviors around 1 Hz are key in control of magnitude distortion, which makes the segment flat or bend in the output.

If using a single 0.3-mV·s pulse test, consider choosing the onset level of the output as the reference point for the 0.1-mV limit, as an alternative way, which is then testable for both linear-phase and nonlinear-phase filters.

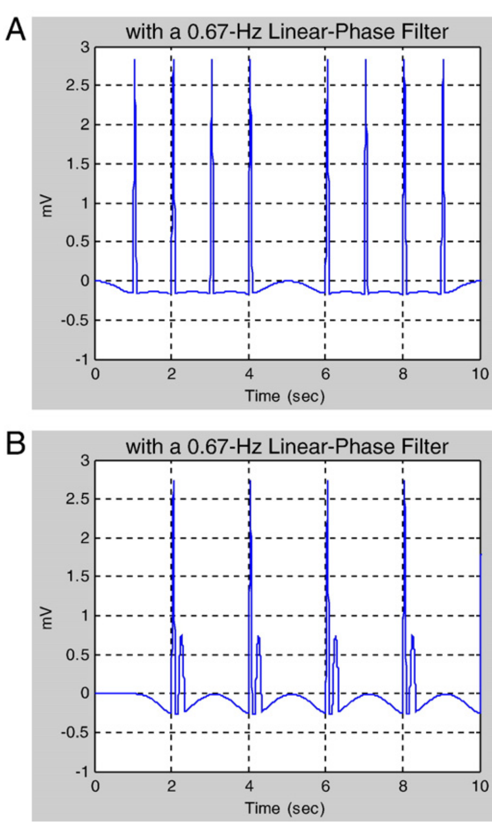

Fig. 6. With a 0.67-Hz linear-phase filter, at 60 cpm rate (A), a pause causes curved baseline segments between 2 triangular pulses, and at 30 cpm rate (B), a QRS-T simulation (from the definitions of test waveforms per AAMI EC13) results in an insignificant impact on ST segments but affects T-P segments.

Line-frequency filter for 50-/60-Hz rejection

Background

Power line frequency interference is a commonly encountered noise contamination in the clinical environment. Data acquisition analog hardware at a very early stage is developed to reduce the line-frequency interference by using common mode rejection circuitry design. However, a certain amount of line-frequency interference voltages still occur in real clinical ECGs.[11] To produce printouts with clean traces, typically, a line-frequency filter (LFF) is activated almost constantly during ECG acquisitions.

An LFF is a band-rejection filter, which passes most frequencies unaltered, but stops the specified band of frequencies from the −3 dB cutoffs f_{c_1} to f_{c_2}. The stop bandwidth (f_{c_2}-f_{c_1}) is typically narrow, as it is also called a notch filter (for an adaptive process and notch filter characteristics, see Appendix III).

An important issue for an LFF in an ECG device is that it can introduce ringing artifacts, an oscillatory behavior, after QRS complexes (Fig. 8). In signal processing, if there is any abrupt bandstop in the frequency domain spectrum, ringing, which is closely related to overshoot and undershoot,

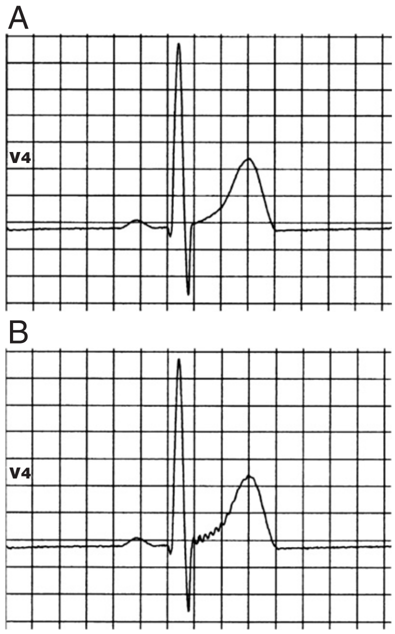

Fig. 8. Notch filter and ringing artifact. The input (A) is analytic wave ANE20000 per IEC 60601-2-51. The output from a notch filter (B) shows the ringing after the QRS complexes. The ringing is due to the abrupt change in the frequency domain spectrum.

appears near the sharp transitions (edges) in the time domain. If the artifact is high, it results in problems such as artificial R wave, uncertain J point and ST segment for detections and measurements.

Compliance review

Currently, there are no specifications for expected variation of the power-line frequency. Correspondingly, we do not see any requirements regarding the stop bandwidth (f_{c_2}-f_{c_1}) for a 50- or 60-Hz LFF. As an extreme case, some basic notch filters in textbook examples[20,21] could have a wide stop bandwidth. If a stop bandwidth is too wide (see Fig. 9), it could be equivalent to a notch plus an LPF. When the notch is on, the QRS heights, for example, could be significantly attenuated.

Regarding the ringing issue, national standards such as AAMI EC11 have no requirements at all. There is nothing to define the stop bandwidth either.

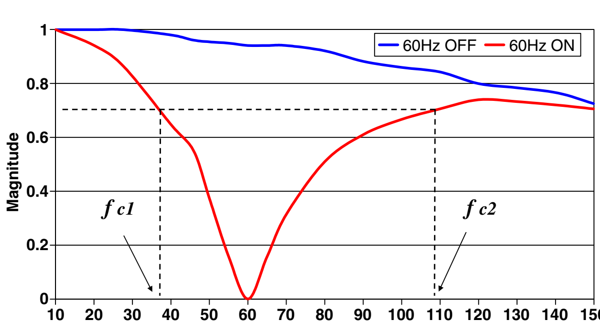

Fig. 9. This LFF notch at 60 Hz has a wider stop bandwidth. Many frequencies are attenuated.

International standard IEC 60601-2-51 has a section (§51.105.3.2) for notch ringing artifact. The test record ECG ANE20000 (Fig. 8a) created by the European CTS-ECG project[11] is artificial data with (*a*) nonzero J-point values in V2, V3, and V4 and (*b*) elevated and sloped ST segments. These 3 leads have larger QRS amplitudes, and the largest ringing artifacts would be mostly visible from them. §51.105.3.2 offers a testing procedure, including preparing 2 printouts with and without notch filter application and comparing the 2 ECG records. The requirement is "Notch FILTERS for line frequency interference suppression shall not introduce on the ECG RECORD more than 25 μV peak ringing NOISE in any LEAD when tested with the test ECG ANE20000." A 25-μV ringing response is only 1/4 mm tall. To measure peak value is technically correct, but practically the comparison is quite difficult to perform even if using a light table. It is partially due to ST-segment morphology of the testing waveform from V2 to V4, and partially because of the trace's line thickness and the small 1/4-mm limit.

Discussion of implications

Because an LFF is often activated in an ECG device, a design with wider stop bandwidth is not suitable as an ECG filter. It is better to provide a specification regarding variation of power-line frequency and correspondingly a narrow range for the stop bandwidth (f_{c_2}-f_{c_1}) for an ECG LFF (and also possibly for harmonic notches, such as 100/120 Hz), especially for interpretive recordings with a passband up to 150 Hz or higher.

The ringing artifact cannot be completely avoided when an LFF is activated. The lack of ringing requirements in the AAMI standards is an important missing item. On the other hand, §51.105.3.2 of IEC 60601-2-51 should be modified so that the notch ringing test is easy to perform. For example, consider using peak-to-valley measurement (not peak value) or pick a different test waveform (with a zero-level J point). Furthermore, the ringing artifact with amplitude decay at the start of a sudden trace change can delay arrival to a desired final

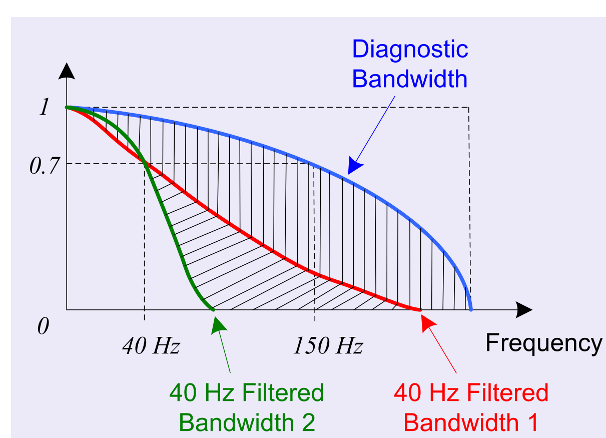

Fig. 10. The diagnostic bandwidth is up to 150 Hz. After 40-Hz filtering, many frequencies are significantly attenuated. Technically, even two 40-Hz LPFs with the same cutoff frequency could have different performances: flatness before f_c (40 Hz) in the magnitude response could affect the amplitude of the QRS, and how fast the drop is after f_c could affect the attenuation of the noise as well as the high-frequency components of the desired signal.

state. Therefore, a requirement of a settling time with error tolerances should be considered. In summary, we should consider updating the standards for ringing requirements and provide a practical guide to evaluate the ringing level.

LPF for muscle artifact and high-frequency interference

Background

For an interpretive ECG device, the AHA/American College of Cardiology (ACC) guidelines today request a passband up to 150 Hz for both adult and pediatric ECGs. Extensive publications have studied higher frequency components (f_c > 150 Hz) for both adults and children,[9,22] and many believe the cutoff should be raised to 250 Hz at least for children[1,23] (here we do not further discuss any cutoffs higher than 150 Hz).

On the other hand, in the clinical environment, ECG printouts and displays compromise diagnostic quality for cleaner traces which results in a "suboptimal" situation, although ECG recordings are sampled and stored with full bandwidth ($f_c \geq 150$ Hz). Most users routinely use artifact filters on ECG printouts, like $f_c = 40$ Hz or even lower, regardless of the manufacturer's workflow design. It is then possible to conclude that only a low percentage of ECGs are actually printed or displayed during overreading at 150-Hz bandwidth. It is well known that an LPF, for example, 40 Hz, could obviously attenuate the QRS complexes and smooth small notches.[9,24] Also, a 40-Hz LPF could filter out the pacemaker spike presentations from ECG printouts.[25] However, for most daily ECG usages, there are no comprehensive studies to examine the diagnostic implications when the bands between 40 and 150 Hz are eliminated (Fig. 10).

Compliance review

Although high-fidelity requirements defined by standards can be available in systems when all filters are off, the actual use of the system with a muscle artifact LPF (MAF) engaged is not well defined. Some guidelines recommend proper labeling for filter impacts and automatically restoring back to 150 Hz for printouts.[1] Currently, there are no printouts from an ECG device specifically alerting the user for possible degraded trace fidelity when commonly used MAFs are activated. If a MAF was enabled, there is no automatic function in most ECG devices to switch back to full bandwidth for the next patient.

Discussion of implications

For compliance, consider an improvement of labeling (a constant alert statement of reduced diagnostic quality) on the printout whenever a MAF is turned on. Also, consider developing an acceptable workflow format for users to add an automatic function to reset back to 150-Hz bandwidth for printouts after a MAF is chosen.

Due to its broad applicability, a study of the 40-Hz LPF regarding clinical significance, incorporating both weak and strong LPF designs (Fig. 10) and their phase responses for filter designs, should be undertaken.

The working groups (AHA/ACC/Common Standards for Quantitative Electrocardiography) should provide a guideline for practicing staff on what impacts to visual and computer diagnostic accuracy will occur when the 40-Hz LPF is activated.

For a much larger task, rules and criteria could be adjusted for the filtered data by extensive database evaluations.

Conclusions

The primary goal of practical filter design is to promote a better signal to noise ratio, including minimizing filter artifacts, such as aliasing, ringing, and segment and amplitude distortions for the passband and transition band. Especially, use linear-phase filters for low-pass and high-pass processing to avoid any potential phase distortion problems across the frequency bands.

An analog LPF should eliminate frequencies near the oversampling frequency. The digital decimation LPFs should reject frequency components at, and greater than, their individual Nyquist frequencies.

An HPF with linear phase at higher cutoff (0.67 Hz) does not cause ST-segment changes referring to the "isoelectric" PR segment, but could introduce T-P segment changes when the heart rate is very low. Consider using the limit of −0.9 dB at 0.67 Hz for interpretive devices and cardiac monitor devices with ST-segment measurements.

The limit of the ringing artifact and stop bandwidth should be well defined, and a test procedure should be easy to perform.

The clinical significance of "suboptimal" conditions when a muscle artifact smoothing filter is activated should be studied due to its extensive usage so that a guideline can be provided for clinical practice.

With respect to compliance, this article has highlighted the fact that existing standards are limited in some areas, and it is suggested that they should be updated at some point in the future.

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Appendix I: Discussion of oversampling and pacemaker spike representation in an ECG printout

Suppose the oversampling rate is 8000 Hz, with the target sampling rate at 500 Hz. We would decimate by a factor of $M = 16$ and use a FIR LPF before down-sampling.

A pacemaker spike is typically 0.5 milliseconds wide on body surface ECGs.[25] An 8000-Hz (0.125-millisecond interval) data stream has 4 data samples to represent a pacer pulse. A FIR filter is in fact a weighted moving average process. To eliminate aliasing, the order (or nonzeros taps) of a proper filter for decimation should be significantly greater than M. Using the present and past data samples within the moving window, therefore, the output of a FIR filtering process can include pacer spike information where the spike datapoints have attenuated amplitudes but possibly are a bit wider and are retained even when the data are down-sampled to 500 Hz afterward.

Appendix II: Discussion of analog LPF, oversampling, and aliasing

If aliasing appears in the passband of desired signals, aliasing interference due to a weak analog LPF cannot be eliminated by postprocessing once the data are sampled or resampled.

Assume ECG bandwidth B_0 = 200 Hz, oversampling frequency f_{os} = 8000 Hz, and target sampling frequency f_s = 500 Hz. Also assume one-stage decimation from f_{os} to f_s and that the digital decimation LPF has strong attenuation at 250 Hz (the target Nyquist frequency $f_s/2$) and above. Further assume a long decreasing transition band from an analog LPF design because the f_{os} is significantly high.

In this case, we thus notice that (*a*) aliasing, from 4000 Hz (the oversampling Nyquist frequency $f_{os}/2$) to 7800 Hz, that is folded outside the ECG passband, can be eliminated by a following digital LPF, and (*b*) aliasing, from 7800 to 8000 Hz, that is folded back into the ECG passband (B_0 to 0), could be at quite a low level after adequate attenuation through the analog LPF.

This analysis shows that aliasing may not be a significant concern even for some worst case if the attenuation from an analog filter at near the oversampling frequency is below a tolerant level (in effect completely rejected).

Second, this analysis tells us some advantages of an oversampling process – we could use a lower order (less stronger) analog LPF where aliasing originating from the wider band will not interfere with the ECG passband. Further, the lower order analog LPF could be designed to have a near linear phase for the ECG passband since the ECG bandwidth is relatively narrow compared to the high oversampling frequency.

Of course, the costs saved from a lower order analog LPF will not be an advantage overall. The decimation process uses FIR filter(s) to get rid of phase distortion. The main disadvantage of FIR filters is that considerably more computation power is required, which will add to hardware costs.

Appendix III: Discussion of adaptive processing and the notch filter characteristics

An adaptive noise canceller is a dual-input system.[26] The 2 inputs are derived from a primary sensor and a reference (or auxiliary) sensor. The primary sensor receives input modeled as the desired signal corrupted by additive noise. The reference sensor receives a noise that is uncorrelated with the signal but correlated with the noise in the primary sensor. Due to the correlated relationship between noises from the primary and reference sensors, the adaptive filter could reject the noise from the primary sensor without altering the desired signal. Widrow et al[27] demonstrated cancelling 60-Hz interference in electrocardiography by using a reference signal taken from a wall outlet. Luo et al[28] illustrated cancelling noise interference in electrocardiography by using a reference signal taken between an additional electrode on the patient's right arm and right arm electrode.

Some ECG devices use an adaptive process to remove the line-frequency interference. Due to practical and economical reasons, a sinusoidal function simulation of 50 or 60 Hz is used for the reference input instead of placing a reference sensor within the noise field of the primary sensor to acquire correlated noise.[29,30] The amplitude and phase of the internal sinusoid are adapted to the powerline noise present in the primary sensor. Mathematically, it is a nonlinear filter[31] and can be identified as an IIR notch filter with nonlinear update processing.[32] Unlike a generic adaptive filter above, this nonlinear filter will "notch" the selected frequency and remove it from both the noise and signal, and therefore, practically this filter requires manual selection for the mains frequency of 50 or 60 Hz.

Like a classical band-rejection filter, we can plot the frequency response by sweeping this nonlinear filter, and the response should demonstrate notch characteristics. Also, both national and international standards continue to call the LFF a notch filter.[11,18,19] Thus, we use the term notch filter in a more generalized sense.

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