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ECG baseline wander removal using the LSRLMF-based fixed-point interference canceller

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***Abstract*—The newly proposed Leaky Sign Regressor Least Mean Fourth (LSRLMF) algorithm is used in a fixed-point interference canceller for ElectroCardioGram (ECG) Baseline Wander (BW) removal application. An upper bound on the step-size of the LSRLMF algorithm is also presented. The LSRLMF-based interference canceller is quantized using various loss of precision methods. Through rigorous simulations, the quantization bit depth required for the different parameters of the LSRLMF-based interference canceller is found to be 9-bits for the most effective ECG baseline wander removal.**

Keywords*—*Baseline wander, bit depth, ECG, fixed-point, LSRLMF.

I. INTRODUCTION

Adaptive filtering algorithms are used for applications such as interference cancellation [1] in physiological signals like ECG. In [2], a total of eight adaptive filtering algorithms were studied for multiple ECG artifacts removal using floating-point interference cancellers. In [3], the classical Least Mean Square (LMS) algorithm was studied for ECG baseline wander removal using fixed-point interference cancellers for the first time.

However, the different variants of the leaky adaptive algorithms [4], [5] have hardly been studied for ECG interference cancellations applications. In [6], [7], two novel leaky adaptive algorithms, viz. Leaky Sign Regressor Algorithm (LSRA) and Leaky Sign Regressor Least Mean Mixed Norm (LSRLMMN) algorithm were used in an interference cancellation application for ECG Power Line Interference (PLI) removal, respectively. In this paper, a new variant of the LLMF algorithm called the Leaky Sign Regressor Least Mean Fourth (LSRLMF) algorithm is used in an interference cancellation application for ECG baseline wander removal.

II. THE LSRLMF-BASED FIXED-POINT INTERFERENCE CANCELLER

The LSRLMF-based fixed-point interference canceller for ECG baseline wander removal is shown in Figure 1. In the figure, the primary input d_i is the ECG signal with baseline wander interference, the secondary input $\mathbf{x}_i$ is the reference baseline wander interference correlated with the baseline wander interference in the primary input, $\mathbf{c}_i$ is the LSRLMF weight vector, y_i is the LSRLMF filter output, and e_i is the corrected ECG signal.

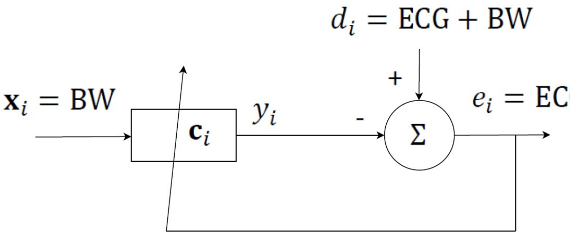

Fig. 1. **The LSRLMF-based fixed-point interference canceller.**

III. UPPER BOUND OF THE LSRLMF ALGORITHM

The LSRLMF weight vector is updated using the below expression:

$$\mathbf{c}_i = (1-\epsilon\beta)\mathbf{c}_{i-1} + \epsilon \, \mathrm{sign}[\mathbf{x}_i]^{\mathrm{T}} e_i^3, \tag{1}$$

where ϵ is the step-size and β is the leakage factor. Subtract Equation (1) from the LSRLMF optimal weight vector $\mathbf{c}^{\mathrm{o}}$ to obtain

$$\widetilde{\mathbf{c}}_i = (1-\epsilon\beta)\widetilde{\mathbf{c}}_{i-1} - \epsilon \, \mathrm{sign}[\mathbf{x}_i]^{\mathrm{T}} e_i^3 + \epsilon\beta\mathbf{c}^{\mathrm{o}}, \tag{2}$$

where $\widetilde{\mathbf{c}}_i = \mathbf{c}^{\mathrm{o}} - \mathbf{c}_i$ is the LSRLMF weight error vector. Apply expectation operator on Equation (2) to obtain

$$\mathrm{E}[\widetilde{\mathbf{c}}_i] = (1-\epsilon\beta)\mathrm{E}[\widetilde{\mathbf{c}}_{i-1}] - \epsilon\mathrm{E}\left[\mathrm{sign}[\mathbf{x}_i]^{\mathrm{T}} e_i^3\right] + \epsilon\beta\mathbf{c}^{o}. \tag{3}$$

It is shown in [8] that

$$\mathrm{E}\left[\mathrm{sign}[\mathbf{x}_i]^{\mathrm{T}} e_i^3\right] = 3\sqrt{\frac{2}{\pi\sigma_x^2}}\sigma_e^2\mathbf{R}\mathrm{E}[\widetilde{\mathbf{c}}_{i-1}], \tag{4}$$

where σ_x^2 is the reference baseline wander variance, σ_e^2 is the corrected ECG signal variance, and $\mathbf{R} = \mathrm{E}[\mathbf{x}_i^{\mathrm{T}}\mathbf{x}_i]$ is the reference baseline wander autocorrelation matrix. Substitute Equation (4) in Equation (3) to obtain

$$\mathrm{E}[\widetilde{\mathbf{c}}_i] = \left[\mathrm{I} - \epsilon\beta - 3\epsilon\sqrt{\frac{2}{\pi\sigma_x^2}}\sigma_e^2\mathbf{R}\right]\mathrm{E}[\widetilde{\mathbf{c}}_{i-1}] + \epsilon\beta\mathbf{c}^{o}. \tag{5}$$

The upper bound of the LSRLMF algorithm as shown in the below expression is obtained from Equation (5):

$$0 < \epsilon < \frac{2\sqrt{\pi\sigma_x^2}}{\beta\sqrt{\pi\sigma_x^2} + 3\sqrt{2}\sigma_e^2\lambda_{\max}}, \tag{6}$$

where $\lambda_{\max}$ is the largest eigenvalue of the reference baseline wander autocorrelation matrix. The expression in Equation (6) is also obtained in a straightforward manner by substituting mixing parameter = 0 in the upper bound expression of the LSRLMMN algorithm in [6].

IV. SIMULATION RESULTS

The ECG signal without baseline wander interference having 3600 samples shown in Figure 2 is derived from the MIT-BIH Arrhythmia Database [9]. It is added with as many samples of baseline wander interference derived from the MIT-BIH Noise Stress Test Database [9] and is shown in Figure 3. The ECG signal with baseline wander interference is passed through five separate LSRLMF-based interference cancellers with the following specifications: weight vector length = 5, reference baseline wander variance = 0.1, step-size = 0.01, leakage factor = 0.002, and number of iterations = 10.

The resulting ECG signals at the adder output of Figure 1 from unquantized, truncate, round, round-to-zero, and convergent round methods used in the respective LSRLMF-based interference cancellers are shown in Figures 4–8, respectively. The resulting ECG signals from unquantized, round, and convergent round methods are found to be corrected for baseline wander interference as shown in Figures 4, 6, and 8, respectively. On the other hand, the presence of baseline wander interference is still evident in the resulting ECG signals from truncate and round-to-zero methods as shown in Figures 5 and 7, respectively. The performance of the unquantized, truncate, round, round-to-zero, and convergent round methods used in the respective LSRLMF-based interference cancellers is also seen in their resulting mean square errors as shown in Figures 9–13, respectively.

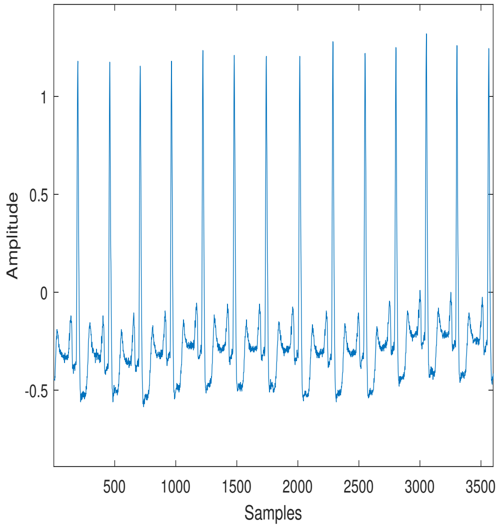

Fig. 2. **ECG signal without baseline wander interference.**

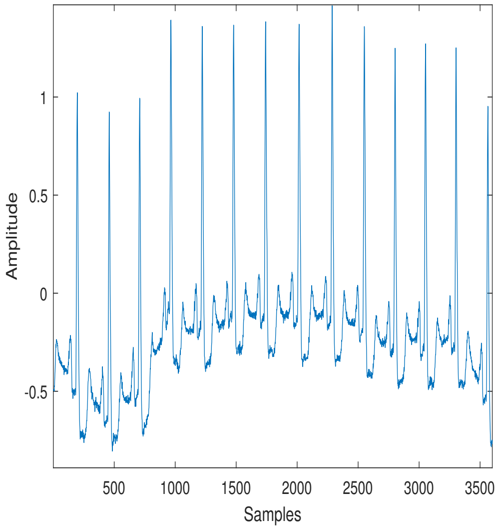

Fig. 3. **ECG signal with baseline wander interference.**

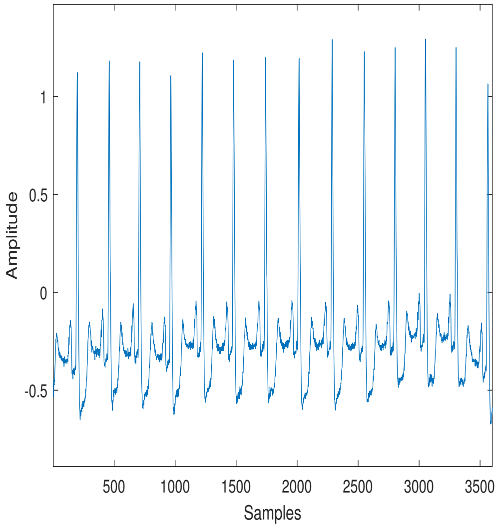

Fig. 4. **Corrected ECG signal from unquantized method using LSRLMF.**

V. CONCLUSION

Using the round and convergent round methods, the quantization bit depth required for the most effective performance of the LSRLMF-based interference canceller for ECG baseline wander removal application is found to be 9-bits for all the parameters, viz. the primary and secondary inputs, step-size, leakage factor, LSRLMF weights, LSRLMF output, and corrected ECG signal. The resultant signal to noise ratios from the round and convergent round methods using LSRLMF is found to be the same i.e., 8.6439 dBs.

ACKNOWLEDGMENT

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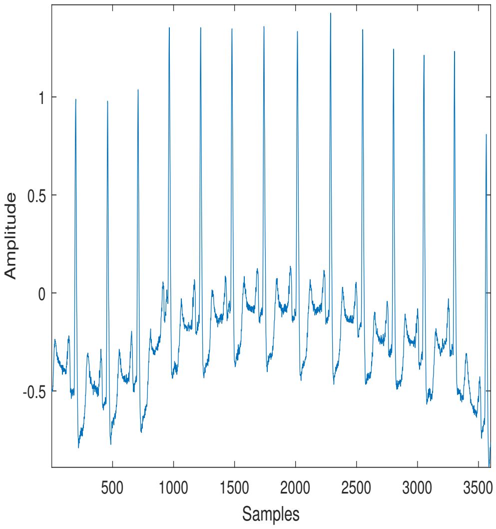

Fig. 5. **Corrected ECG signal from truncate method using LSRLMF.**

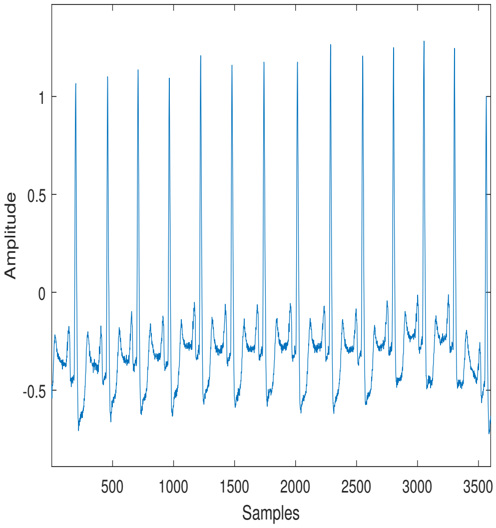

Fig. 6. **Corrected ECG signal from round method using LSRLMF.**

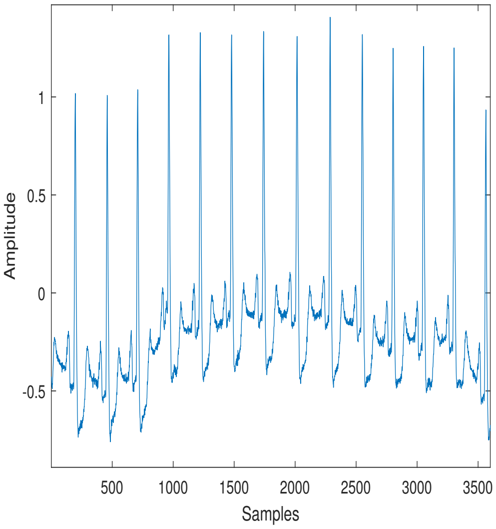

Fig. 7. **Corrected ECG signal from round-to-zero method using LSRLMF.**

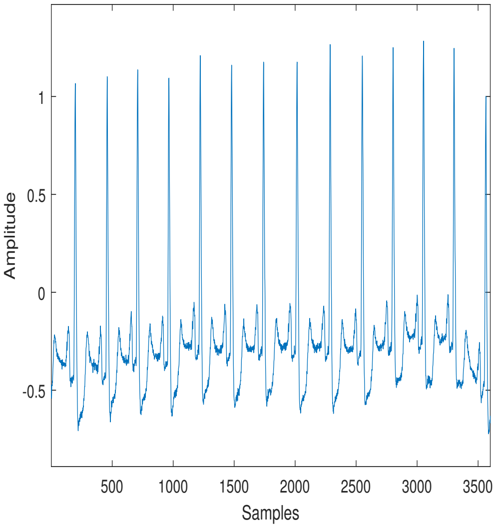

Fig. 8. **Corrected ECG signal from convergent round method using LSRLMF.**

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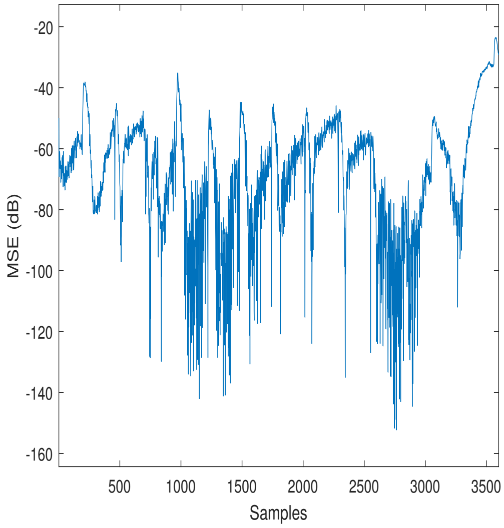

Fig. 9. **Resultant mean square error from unquantized method using LSRLMF.**

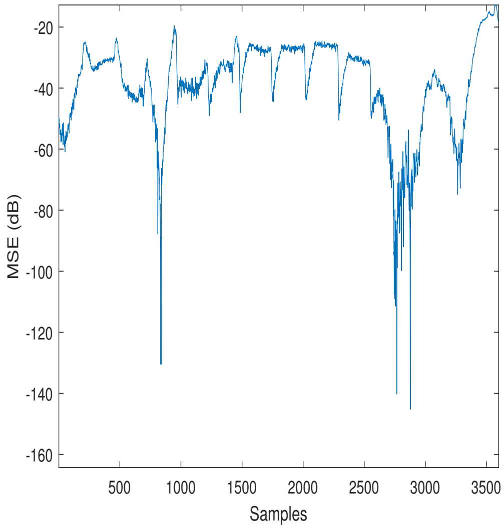

Fig. 10. **Resultant mean square error from truncate method using LSRLMF.**

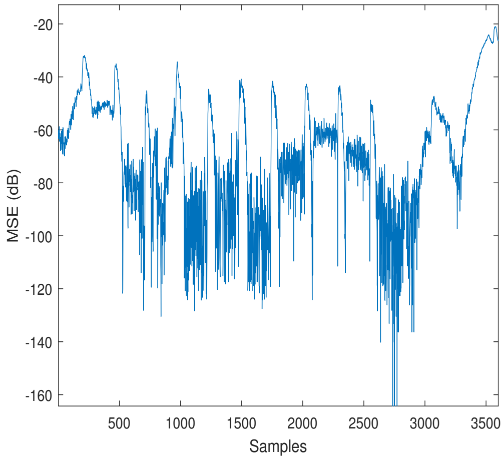

Fig. 11. **Resultant mean square error from round method using LSRLMF.**

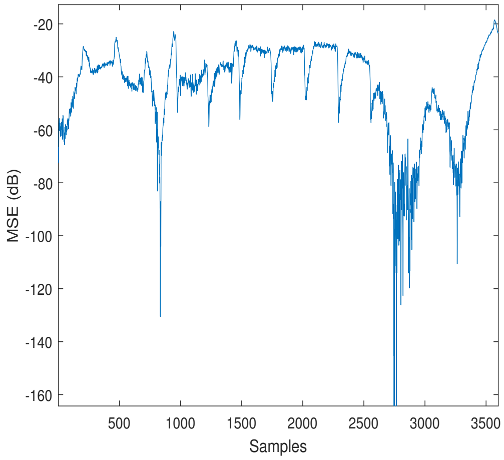

Fig. 12. **Resultant mean square error from round-to-zero method using LSRLMF.**

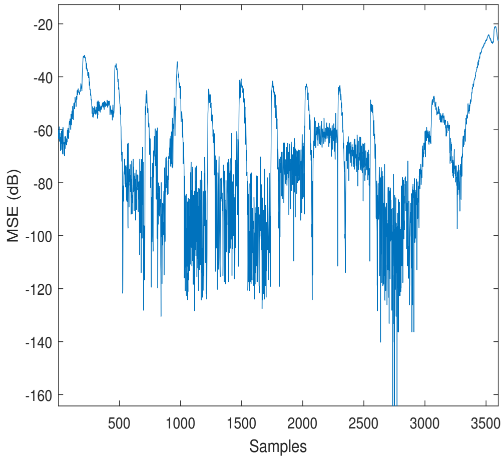

Fig. 13. **Resultant mean square error from convergent round method using LSRLMF.**