A Medical-Grade Wireless Architecture for Remote Electrocardiography

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***Abstract*—In telecardiology, electrocardiogram (ECG) signals from a patient are acquired by sensors and transmitted in real time to medical personnel across a wireless network. The use of IEEE 802.11 wireless LANs (WLANs), which are already deployed in many hospitals, can provide ubiquitous connectivity and thus allow cardiology patients greater mobility. However, engineering issues, including the error-prone nature of wireless channels and the unpredictable delay and jitter due to the nondeterministic nature of access to the wireless medium, need to be addressed before telecardiology can be safely realized. We propose a medical-grade WLAN architecture for remote ECG monitoring, which employs the point-coordination function (PCF) for medium access control and Reed–Solomon coding for error control. Realistic simulations with uncompressed two-lead ECG data from the MIT–BIH arrhythmia database demonstrate reliable wireless ECG monitoring; the reliability of ECG transmission exceeds 99.99% with the initial buffering delay of only 2.4 s.**

***Index Terms*—Electrocardiogram (ECG), IEEE 802.11, telecardiology, wireless healthcare.**

I. Introduction

Wireless telecardiology is one of the most promising examples of telemedicine, which involves the real-time transmission of electrocardiographic signals over wireless networks [1]–[4]. Among several candidates for wireless electrocardiogram (ECG) monitoring, the IEEE 802.11 wireless LAN (WLAN) is a promising solution, which is already being deployed in many hospitals [5]–[10].

IEEE 802.11 WLAN technology is simple, flexible, and cost-effective, makes it suitable as a ubiquitous communication environment in hospitals. However, safety-critical medical applications require the strict maintenance of a particular quality of service (QoS), which includes the provision of a minimum bandwidth and delay, and a maximum jitter and error rate [11]. Guaranteeing those requirements using an IEEE 802.11 WLAN is challenging because wireless channels are subject to errors, and the medium access control (MAC) of IEEE 802.11 is not designed to assure a strict QoS. Moreover, a high collision rate and the resulting requirement for frequent retransmissions can cause unpredictable delays and jitters in a wireless network, which degrade the quality of real-time ECG services.

In this paper, we first identify and analyze the errors introduced during transmission over a wireless channel communicating with mobile devices using both quantitative and qualitative criteria. Second, we present a *novel*, medical-grade WLAN architecture, in which the IEEE 802.11 MAC layer is split into MAC and logical-link-control (LLC) layers, in order to achieve a QoS suitable for real-time telecardiology. The MAC layer operates in point-coordination-function (PCF) mode [12], which uses a round-robin scheduler to poll each station (STA) sequentially. Additionally, the LLC layer incorporates an error-control structure which is designed to achieve the dependability required for ECG monitoring by combining an forward error correction (FEC) technique with block interleaving, so as to achieve homogeneous throughput with a bounded delay. Finally, we evaluate the effectiveness of the proposed architecture in achieving reliable ECG monitoring services using ECG recordings from the MIT–BIH database [13], [14]. Although no real-time patient data in situation is being used, this allows us to form a realistic assessment of the reliability of the proposed novel wireless architecture in the context of the telecardiology application.

The main contribution of our work is to prevent collision between STAs by introducing enhanced QoS-aware MAC coordination. This allows data to be delivered in a deterministic way, and thus meets the jitter constraints of telecardiology. In addition, the proposed LLC, in combination with the enhanced MAC coordination, is shown to promote the reliability of ECG data transmission while guaranteeing a bounded delay. Our early-phase QoS assessment of the new wireless architecture is essential because ECG applications are safety-critical, and failure after deployment can have serious consequences.

The rest of this paper is organized as follows. In Section II, we present the background to the use of a IEEE 802.11 WLAN for telecardiology. In Section III, we introduce a QoS enhancement architecture for IEEE 802.11, focusing on the MAC layer. In Section IV, the focus switches to the LLC layer. Our evaluation environment is introduced in Section V, and then, we present results to show the effectiveness of the proposed architecture in providing improved QoS. Related studies of ECG monitoring

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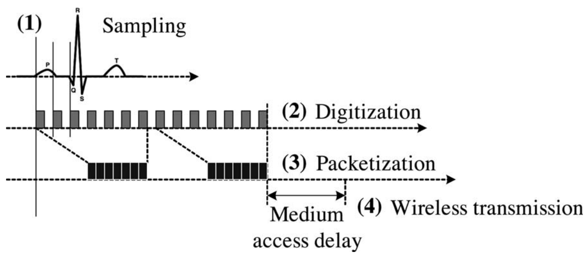

Fig. 1. Sampling, digitization, and packetization of ECG data for transmission over an IEEE 802.11 WLAN.

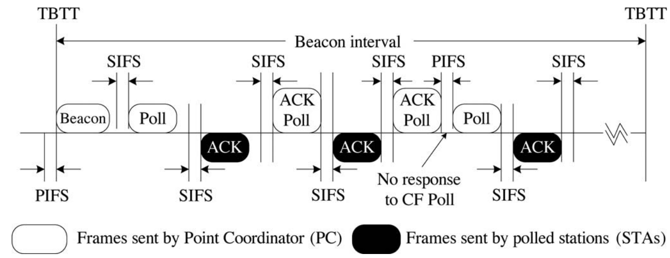

Fig. 2. Exchange of frames between STAs using the PCF-based MAC protocol to guarantee predictable delivery of ECG data.

are reviewed in Section VI. Finally, Section VII is devoted to conclusions.

II. ECG Transmission Over IEEE 802.11 WLANs

Electrocardiography is a noninvasive technique that measures the electrical activity of the heart. A typical ECG trace consists of P-wave, QRS complex, and T-wave [6] signals that are periodically sampled (see (1) in Fig. 1) by electrodes attached to the patient, and then, digitized (see (2) in Fig. 1). The sampling frequency and digital resolution determine the characteristics of the wireless traffic generated by the subsequent transmission. There are a number of possible sampling and digitizing methods [3], [4], [15], as well as various compression techniques [16]–[18].

The number of leads required depends on the particular ECG configuration. If there are N_r leads, and the signal from each lead is digitized at a rate of N_s samples per second with a resolution of L_s bits, then the resulting data rate of the ECG application would be

$$\mu_{\mathrm{ECG}} = N_r N_s L_s. \tag{1}$$

This stream of digital data is packed into frames in the packetization process (see (3) in Fig. 1), and then may be sent to a remote monitoring device across a wireless channel. Fixed access points (APs), attached to a wired network infrastructure, provide a communication portal for all the STAs within its range. The ECG data are transmitted to the AP, and then relayed to a remote monitoring device through the hospital LAN. This wired infrastructure can naturally be expected to deliver data much faster and more reliably than a wireless network, and so we can attribute all data losses to the wireless network without significant inaccuracy.

There are two main trends in the deployment of wireless ECG monitoring: a vendor-specific network which uses dedicated Wireless Medical Telemetry Service (WMTS) bands, and the IEEE 802.11 WLAN which uses shared Industrial, Scientific and Medical (ISM) bands [9]. While a vendor-specific network in the WMTS bands enjoys the benefit of dedicated bands, it suffers from low bandwidth. On the contrary, an IEEE 802.11 WLAN has benefit in cost by the standard-based deployment as well as its large bandwidth. However, because the ISM bands are unlicensed and are subject to interference from sources such as Bluetooth devices and microwave ovens. Nevertheless, experience based on recent substantial deployments suggests that an IEEE 802.11 WLAN can significantly outperform the conventional vendor-specific one [9]. That is why we focus on the design of an efficient IEEE 802.11 WLAN architecture which can guarantee the QoS level required for telecardiology.

For multiple channel access, the IEEE 802.11 standard defines two medium-access coordination functions: the basic distributed coordination function (DCF) and the optional PCF [12]. DCF is a distributed medium-access scheme based on a carrier-sense multiple-access with collision avoidance (CSMA/CA) protocol. In DCF mode, an STA must sense the medium before initiating a packet transmission, and must then wait for an exponential random back-off period if contention occurs between different STAs trying to access the medium. This is likely to happen because all the STAs in one basic service set (BSS[1]) compete for the channel with the same priority. This makes it impossible to guarantee any forms of QoS which depends on an assured bandwidth, packet delay, or jitter.

Although there has been a good deal of work on telecardiology [1]–[4], [6]–[8], to the best of our knowledge, there have been no previous attempts to solve the problem of guaranteeing QoS for ECG transmissions over a WLAN with the effective combination of medium access and error control schemes. We will now address this issue.

III. Medical-Grade MAC in IEEE 802.11 WLAN for Delay Guarantee

In order to guarantee predictable delivery of ECG data, most of all, we should design an appropriate MAC protocol. We advocate the use of PCF for ECG data transmission. PCF uses a centralized polling scheme, which requires the AP to act as a point coordinator (PC). If a BSS is set PCF-enabled, then its channel access time is divided into beacon intervals, as shown in Fig. 2. A beacon interval is composed of a contention-free period (CFP) and a contention period (CP). During the CFP, the PC maintains a list of registered STAs, which are polled in the order in which they are listed. When an STA is polled, it receives permission to transmit a data frame. Since every STA is allocated a maximum length of frame to transmit, the maximum duration of the CFP for all the STAs can be determined by the PC. The time taken by the PC to generate beacon frames is called the target beacon transmission time (TBTT). The PC specifies the next TBTT within the beacon, which is broadcast to all the other STAs in the BSS. In a nutshell, due to the round-robin scheduler, each STA in PCF is guaranteed to transmit at least once in each cycle.

[1] A group of STAs coordinated by DCF or PCF is called a BSS.

A typical medium-access sequence using PCF is shown in Fig. 2. When a PC polls an STA, it can piggyback the data-frames to the STA together with the CF-poll frame. After a short interframe space (SIFS) interval the STA then sends back a data frame piggybacked with an acknowledgement (ACK). When the PC polls the next STA, it piggybacks not only the data-frame to its destination, but also an ACK to the previous successful transmission. Note that packet transmissions are separated by SIFS except in one scenario: if the polled STA does not respond to the PC within the period of a PCF interframe space (PIFS), the PC will poll the following STA. Silent STAs are removed from the polling list after several periods, but may be polled again at the beginning of the next CFP. Normally, PCF uses a round-robin scheduler to poll each STA sequentially in the order of a polling list, but priority-based polling mechanisms can also be used if different QoS levels are required by different STAs.

IV. Medical-Grade LLC for Robust ECG Transmission over IEEE 802.11 WLANs

We introduce the LLC layer above the MAC layer to enhance the QoS for ECG applications. Proper error control is required at the LLC layer in the subnetwork because of the errors that occur in the wireless link. However, retransmitting information is not appropriate for real-time ECG applications because of the nondeterministic delay that takes place during error recovery. On the contrast, the FEC approach has a homogeneous throughput and a bounded time delay, which are crucial considerations for ECG applications.

We advocate the use of Reed–Solomon (RS) coding, combined with interleaving [19], as a method of FEC in the LLC layer. We use a simple RS code with short codewords, with the aim of reducing the buffering delay to a level that is compatible with the moderate data rate of ECG applications.

1) RS FEC Codes: The proposed error control scheme introduces RS FEC codes in the LLC layer (between (3) and (4) in Fig. 1), which we have chosen because of their superior performance at lower error rates [20].

An RS code is specified by (n, k) with s-bit symbols, which takes k data symbols of s bits each and adds $n - k$ parity symbols to make an n-symbol codeword. An algebraic RS decoder can correct up to t symbols that contain errors in a codeword or up to $2t$ erasures, where $2t = n - k$. We use the RS code as an erasure code (not as an error-correcting code), and assume that the cyclic redundancy check (CRC) provided by the physical layer will be used to detect and erase damaged physical-layer frames. Erasure decoding has the advantage that it is simpler than error correction, because the position of the erased octets is known in advance.

RS decoding is performed for each codeword. When a codeword is received, a syndrome symbol is created for each parity symbol [21]. If there are any errors in that codeword, then their locations are found using the Berlekamp algorithm [22]. The original data can then be recovered by erasure decoding. The number of times that the decoding procedures need to be executed is proportional to the number of errors, which may occur in up to t symbols. Thus, if we assume that $T_{\mathrm{cw}}(\nu)$ is the time required for decoding a single codeword which contains ν symbol errors, then $T_{\mathrm{cw}}(\nu)$ also bounds the delay incurred in the RS decoding of a codeword on $T_{\mathrm{cw}}(t)$, and it is this bound that makes the decoding process predictable for real-time ECG applications.

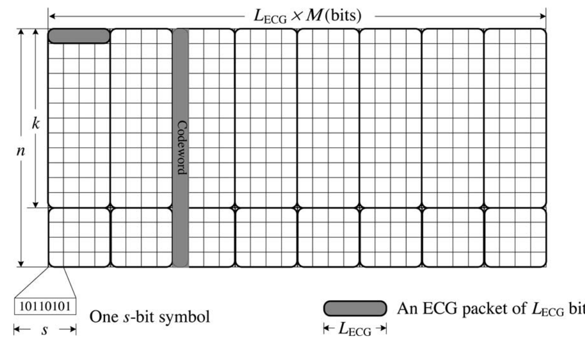

Fig. 3. 2-D array buffer used to apply block interleaving to RS coding.

2) Interleaving to Combat Error Bursts: FEC incurs constant transmission overhead even when the channel is error free. To ensure data fidelity while minimizing the transmission overhead, we adopted an interleaving technique [19]. The block interleaving [23], first creates a 2-D array buffer like the one shown in Fig. 3. The ECG packets, each of which is L_{ECG} bits, are read into the 2-D array in rows, and RS coding is applied along columns. Then, the data are pushed out in rows to the physical layer for transmission over the physical wireless channel. Now, the error bursts encountered during transmission are spread across multiple codewords. Consequently, the number of errors occurring within one codeword may be sufficiently small to allow them to be corrected using a simple FEC technique.

The value of M which is the number of ECG packets in each buffer row, determines the degree of interleaving. A high value of M provides increased time diversity, which improves performance in the presence of time-varying shadowing, at the cost of a larger buffer and a longer buffering delay. The amount of storage required is nML_{ECG} bits, and the corresponding buffering delay δ is determined as follows:

$$\delta = \frac{nML_{\mathrm{ECG}}}{\mu_{\mathrm{ECG}}} \tag{2}$$

where μ_{ECG} is the data rate of the ECG application.

V. Evaluation of the Proposed QoS Enhancement Architecture

A. Error Statistics for Packet Transmissions Over a Fading Channel

Unlike wired communication, channel errors are common in wireless networks. A typical bit error rate is 10^{-5}, leading to a packet error rate (PER) of a few percents; however, a PER can even go up to 10%, depending on other parameters such as the transmission power and the distance between the transmitter and receiver. Even worse, errors in a wireless network tend to occur in bursts due to the nature of the fading channel.

The binary error process that describes the success or failure of data block transmission has been considered by Zorzi *et al.* [24], where the error process is specified by two independent parameters q and r, where q is the probability that the transmission of the ith block is unsuccessful, given that the $(i-1)$th block was transmitted successfully, while r is the probability that the ith block is successfully transmitted, given that the $(i-1)$th block was not. Using these parameters, the analytical expression for the Markov parameters for Rayleigh fading is given as follows:

$$\epsilon = 1 - e^{-1/F} = \frac{q}{q+r} \tag{3}$$

where ϵ is the steady-state PER and F is the fading margin. Now, the value of r can be calculated as follows:

$$r = \frac{Q(\theta, \rho\theta) - Q(\rho\theta, \theta)}{e^{1/F} - 1} \tag{4}$$

where $Q(\cdot,\cdot)$ is the Marcum-Q [25] function, and

$$\theta = \sqrt{\frac{2/F}{1-\rho^2}} = \sqrt{\frac{-2\log(1-\epsilon)}{1-\rho^2}}. \tag{5}$$

The term ρ is the correlation coefficient of two samples of the complex Gaussian fading process, and is equal to $J_0(2\pi f'_D)$, where $J_0(\cdot)$ is the Bessel function of the first kind and of zeroth order, and f'_D is the normalized version of the Doppler frequency. In particular, the value of f'_D is calculated as $f'_D = f_D L_{\text{phy}}/\mu_p$, where f_D is the maximum Doppler frequency, L_{phy} is the frame size, and μ_p is the reference channel data rate. The maximum Doppler frequency of the system f_D is given as $f_c v/c$, where v is the mobile speed, c is the speed of the electromagnetic wave, and f_c is the carrier frequency.

B. Simulation Environments

1) ECG Databases: We used the MIT–BIH arrhythmia database [13], [14] to evaluate our wireless architecture, which is widely used to test the performance of ECG transmission [3], [4]. This database contains 48 half-hour excerpts of two-channel (MLII[2] and V5[3] channels) ambulatory ECG recordings, obtained from 47 subjects studied by the BIH Arrhythmia Laboratory. Although the database was originally created as standard test material for evaluation of arrhythmia detectors, it is by far the most widely used data for testing and comparing proposed strategies for real-time ECG services. The recordings were digitized at 360 samples per second per channel, with 11-bit resolution over a 10 mV range, and thus the data rates of both the MLII channel μ_{II} and the V5 channel μ_{V5} are 3960 bits/s. Data packets from both the MLII and V5 channels are multiplexed before transmission, and the resulting data rate μ_{ECG} is 7920 $(2 \times 360 \times 11)$ bits/s. We assume that the ECG data are transmitted without compression.

[2]Lead II is the voltage between the (positive) left leg electrode and the right arm electrode.

[3]One of the precordial leads that is placed directly on the chest.

TABLE I

VALUES OF THE EXPONENT h CORRESPONDING TO THE iTH ROW AND jTH COLUMN OF THE PARITY MATRIX FOR THE RS BLOCK CODE (16, 12), SUCH THAT $p_{i,j} = \alpha^h$, WHERE α IS AN ELEMENT OF GF(2^8)

Row index i	$p_{i,12}$	$p_{i,13}$	$p_{i,14}$	$p_{i,15}$
0	40	138	141	8
1	8	196	97	158
2	158	4	250	209
3	209	123	27	76
4	76	226	198	160
5	160	142	95	125
6	125	19	59	70
7	70	87	39	137
8	137	169	244	254
9	254	192	27	160
10	160	57	53	201
11	201	246	201	0

2) RS Code Parameters: An RS code with short codewords is desirable for ECG applications with a moderate data rate, because a short code limits the delay incurred by initial buffering. The error control parameters that we use are inspired from the design of the FEC process for the CDMA2000-1x broadcast air specification [26].

An RS code uses 8-bit symbols and operates in the Galois Field called GF(2^8). The primitive element α for this field is defined by

$$\alpha^8 + \alpha^4 + \alpha^3 + \alpha^2 + 1 = 0. \tag{6}$$

The jth code symbol v_j $(j = 0, \ldots, n-1)$ is then defined by

$$v_j = \begin{cases} u_j, & 0 < j \le k-1 \\ \sum_{i=0}^{k-1} u_i p_{i,j}, & k \le j \le n-1 \end{cases} \tag{7}$$

where u_j is the jth of a block of k information symbols, and $p_{i,j}$ is the entry on the ith row and the jth column in the parity matrix of the code. Table I lists the power h of the entry on the ith row and the jth column of the parity matrix for the (16, 12) RS code, $p_{i,j} = \alpha^h$, where α is a primitive element of GF(2^8), $i = 0, \ldots, 11$, and j = 12, 13, 14, and 15. For example, the entry of 40 in the upper left-hand corner indicates that $p_{0,12} = \alpha^{40}$.

An RS code specified by (16, 12) generates sixteen code symbols for each block of 12 information symbols input to the encoder: the remaining four symbols are parity symbols. The generator polynomial for the (16,12) code is

$$g(X) = 1 + \alpha^{201}X + \alpha^{246}X^2 + \alpha^{201}X^3 + X^4. \tag{8}$$

3) Other Simulation Parameters: Because we used two-channel ambulatory ECG recordings, two buffer blocks, shown in Fig. 3, are required to perform RS encoding at the sending STA (equivalent to patient-worn medical device), and these two buffer blocks are multiplexed before transmission on the physical layer. More specifically, each row of the buffer block forms the payload of an ECG packet, either for the MLII or the V5 channel. The packets are transmitted in physical-layer frames assigned to each channel in the same order as the start times of the transmissions of the physical-layer frames. We assume that this transmission involves the use of a multiplexer (MUX) and

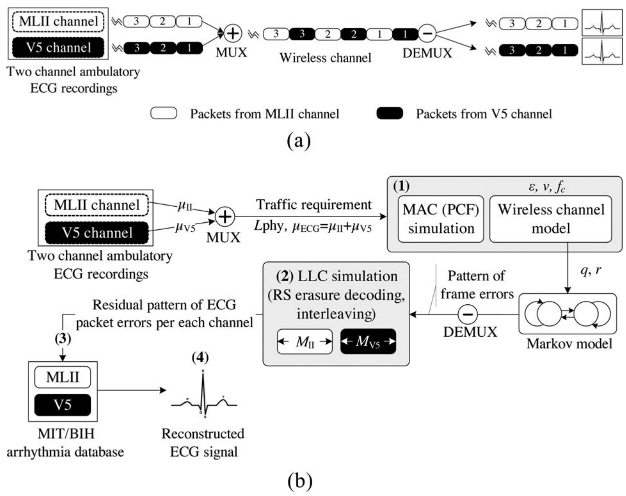

Fig. 4. Simulation structure for QoS evaluation in telecardiology. (a) MUX and DEMUX. (b) High-level simulation architecture.

TABLE II
SIMULATION PARAMETERS FOR ASSESSMENT OF ECG TRANSMISSION USING THE PROPOSED WIRELESS ARCHITECTURE

Symbol	Value(s)	Description
$L_{\rm phy}$	400 bits	Length of a physical-layer frame
$L_{\rm ECG}$	396 bits	Length of a MAC-layer packet
f_c	2.4 GHz	Carrier frequency
μ_p	1 Gb/s	Reference channel data-rate
ϵ	$0 \sim 0.03$	Steady-state PER
v	2∼5 km/h	Mobile speed of patients
s	8 bits	Length of a symbol
(n, k)	(16,12)	Candidate RS code
N_r	2	Number of leads
N_s	360 Hz	Samples per second
L_s	11 bits	Sample size
$\mu_{\rm II}$, μ_{V5}	3960 b/s	Data-rate of each ECG recording channel
$\mu_{\rm ECG}$	7920 b/s	Total ECG data-rate

demultiplexer (DEMUX), as shown in Fig. 4(a). The width of buffer blocks for each channel is $M_{\rm II} L_{\rm ECG}$ and $M_{\rm V5} L_{\rm ECG}$ bits, respectively. The values of $M_{\rm II}$ and $M_{\rm V5}$ are assumed to be less than or equal to 8.

Because hospital patients can be expected to have low mobility, we assumed a speed v of between 2 km/h and 5 km/h. We also varied the PER experienced by the ECG sensors attached to a patient between 0 and 0.03. Then, we analyzed the reliability of ECG transmission using the proposed wireless architecture for different amounts of interleaving. All the parameters used in our simulation are summarized in Table II.

Error traces to model the fate of frames transmitted over the wireless channel were obtained by simulating the PCF in the MAC layer and the channel, using the model explained in Section V-A (see (1) in Fig. 4(b)). The traces were then supplied to the RS erasure decoding simulator (see (2) in Fig. 4(b)). All the simulations were implemented in C, and compiled and run under Linux. This simulation generated residual ECG packet errors and their locations after RS decoding and these were injected into the MIT–BIH arrhythmia data that we used (see (3) in Fig. 4(b)). Finally, the reconstructed ECG signals were

TABLE III
RELIABILITY OF AN ECG MONITORING SERVICE USING THE PROPOSED WIRELESS ARCHITECTURE, EXPRESSED AS THE PROPORTION OF SUCCESSFUL PACKET TRANSMISSIONS OVER THE MLII CHANNEL

$M_{\rm II}$	1	2	4	8
Residual rate of packet errors	3.8E-3	5.9E-4	5.3E-5	3.0E-5
Reliability	99.62%	99.94%	99.99%	99.99%

evaluated (see (4) in Fig. 4(b)) by determining the mean-square errors (MSEs), which quantify the difference between the reconstructed ECG signals and the original signals obtained from the patient. The MSE for the time interval t can be estimated as follows:

$$\mathrm{MSE}_t = \frac{1}{N_s t} \sum_{i=1}^{N_s t} (s_i - \hat{s_i})^2$$

where s_i is the digitized value of the ith ECG signal obtained from the patient and $\hat{s_i}$ is the digitized value of ith ECG signal reconstructed at the remote monitoring device after wireless transmission.

C. Simulation Results

Table III shows the upper bound on the residual rate of ECG packet errors after RS decoding for the MLII channel with different values of the block interleaving parameter $M_{\rm II}$, under the range of channel conditions we assumed. We see that the proposed wireless architecture is sufficiently reliable for remote ECG monitoring. Especially, a high $M_{\rm II}$ value provides increased time diversity and reduces the perceived error rate in the presence of time-varying shadowing. For example, choosing a value of $M_{\rm II}$ that is larger than 4 can provide a reliability of 99.99%.

Figs. 5 and 6 show how the residual rate of ECG packet errors directly affects the quality of ECG signals reconstructed at the remote monitoring device for the selected "100.dat" data stream. Compared to the original ECG signal in Fig. 5(a), the reconstructed ECG signal with no error control in the LLC layer shows that ECG signals are frequently missed; this might lead a physician to misinterpret a patient's condition. For example, the original ECG signal has 13 QRS complexes, while the ECG signal without error control has only 11 QRS complexes, as shown in Fig. 5(b). The missing QRS complexes result in a difference in the RR interval, as compared to the original one. As a result, in spite of the fact that the original ECG diagnosis is normal sinus rhythm with an atrial premature beat, the distorted reconstruction leads to a diagnosis that indicates sinus pause or sinoatrial block, which is a more serious problem. What is even worse, life-threatening arrhythmia, which is very infrequent can be overlooked. A similar phenomenon can also be found in the case of the ECG signal obtained from the V5 channel, as shown in Fig. 6(b).

However, we can see from Figs. 5 and 6 that the ECG signals seen by a physician will be less distorted by the occurrence of frequent error bursts during transmission when the proposed error control in the LLC layer is applied and as the level of

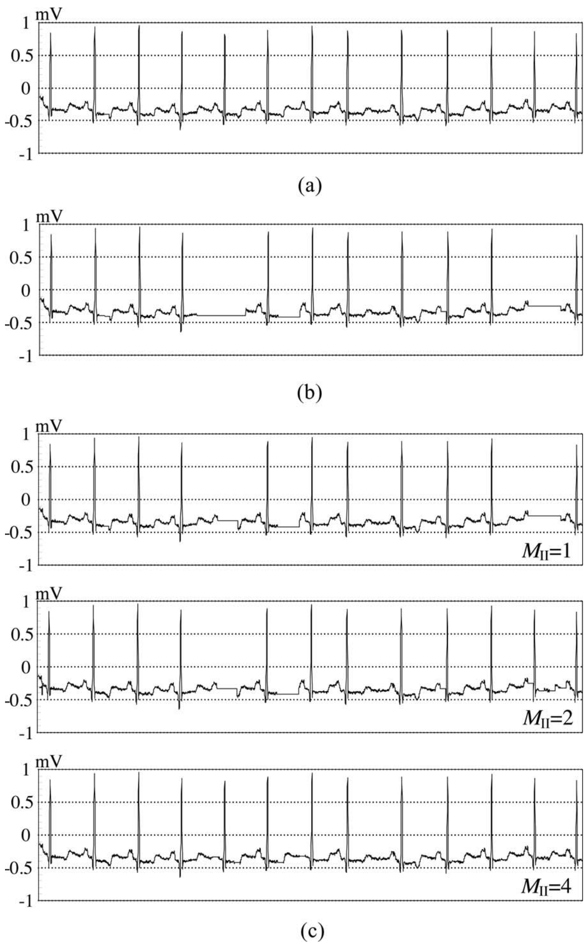

Fig. 5. Analysis of an ECG signal (lead MLII) over 10 s, for a selected interval that contains a lot of error bursts ($v = 2$ km/h, PER $= 0.03$). (a) Original ECG signal sent by patient's STA. (b) Reconstructed ECG signal with no error control in the LLC layer. (c) Reconstructed ECG signal with the proposed error control in the LLC layer for different values of M.

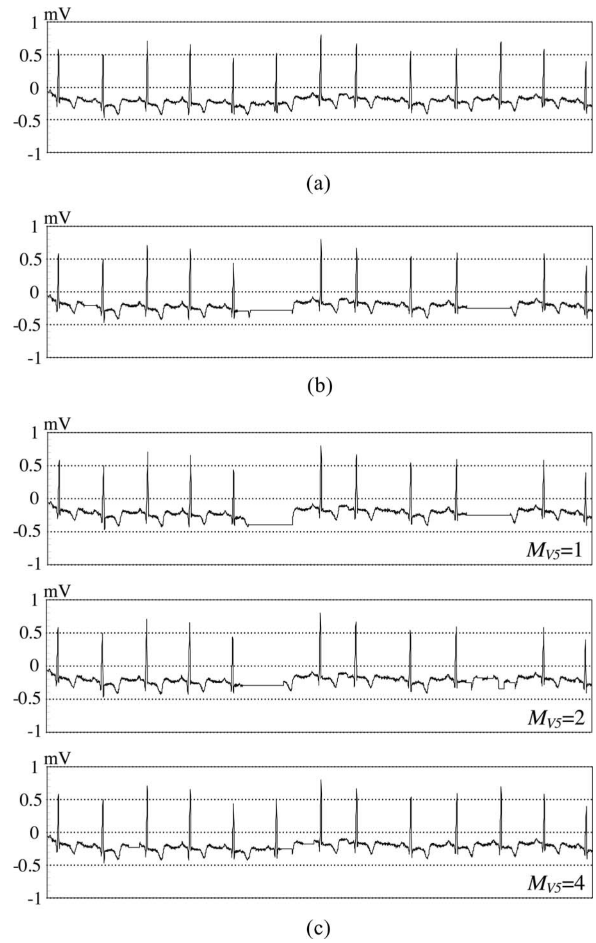

Fig. 6. Analysis of an ECG signal (lead V5) over 10 s, for a selected interval that contains a lot of error bursts ($v = 2$ km/h, PER $= 0.03$). (a) Original ECG signal sent by patient's STA. (b) Reconstructed ECG signal with no error control in the LLC layer. (c) Reconstructed ECG signal with the proposed error control in the LLC layer for different values of M.

block interleaving increases. The reconstructed signal becomes increasingly similar to the original ECG signal as a sufficient level of interleaving is adopted, thus enabling cardiologists to make a correct diagnosis. When the value of M_{II} is greater than 4, the reconstructed ECG signal is almost identical to the original ECG signal. Table IV clearly demonstrates the fact that the values of MSEs at half-hour intervals decrease significantly as the value of M increases; these values have been obtained for ten selected subjects from the MIT–BIH arrythmia database. In order to further demonstrate the performance of the proposed scheme, we obtained the MSEs for the entire dataset of 47 ECG recordings. The results can be seen in Fig. 7, which shows the maximum, minimum, and average MSEs for each case. These results provide further evidence that the proposed LLC structure reduces wireless channel errors in a very effective manner. However, this improvement in error rate comes at the cost of an increased buffering delay. This is 0.6, 1.2, 2.4, and 4.8 s when the value of M_{II} is 1, 2, 4, and 8 respectively. So it becomes important to find the right balance between the quality of the reconstructed ECG signal and the buffering delay.

TABLE IV
MSES OF RECONSTRUCTED ECG SIGNALS USING TEN SELECTED AMBULATORY ECG RECORDINGS WHEN $v = 2$ KM/H AND PER IS 0.03 (MLII CHANNEL)

Record #	Without LLC	With LLC (M)		
		1	2	4
100	5.10E-5	4.51E-5	2.65E-5	1.31E-6
105	2.51E-4	2.12E-4	1.84E-4	1.93E-5
111	6.91E-5	5.13E-5	1.35E-5	3.12E-6
115	1.40E-4	9.11E-5	4.58E-5	7.77E-6
120	2.36E-4	1.09E-4	3.52E-5	6.24E-6
200	1.71E-4	1.78E-4	5.29E-5	5.13E-6
205	7.83E-5	6.15E-5	1.34E-5	1.92E-6
210	1.94E-4	1.77E-4	5.46E-4	1.08E-5
215	3.17E-5	3.04E-5	4.51E-5	1.03E-6
220	8.74E-4	5.96E-4	4.24E-4	2.37E-5
Average	1.98E-4	1.55E-4	1.38E-4	8.01E-6

VI. RELATED WORK

Real-time cardiac monitoring can provide patients with more freedom, but its acceptability relies on the integration of new technologies such as wireless sensors and real-time automatic ECG diagnosis into cardiac monitoring systems. We will briefly mention some existing wireless ECG monitoring systems: HP's Agilent telemetry system is expensive and is generally used for

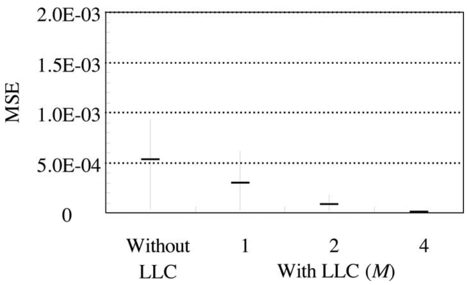

Fig. 7. MSEs with and without the proposed LLC layer, for the entire dataset of 47 ECG recordings. The maximum, minimum, and average MSEs are shown in each case.

multi-patient hospital applications [27]. In this system, patients must stay in hospital for cardiac surveillance.

Braecklein *et al.* [28] implemented a telecardiological monitoring system in which ECG signals are collected and analyzed by wireless ECG sensors. Detected cardiac events are automatically transmitted to the local base station, from whence they travel via a modem and dedicated telephone line to an Internet-based electronic health record (EHR) where the ECG data and event markers are stored. Authorized staff have access to the EHR to read the patient's files. It can transmit a 1-lead ECG signal sampled at 500 Hz and provide real-time transmission of an ECG signal.

A new Australian mobile-phone-based medical diagnostic system called ®LifeMedic has been used to give medical services to the survivors in the region of Banda Aceh, Indonesia, devastated by the tsunami of January 2005 [29]. LifeMedic, which was developed by a Brisbane-based company, can deliver patient care in a hospital or at a remote location. Patient data, both signals and images, are provided respectively by medical sensors (ECG electrodes) and digital cameras, and then transmitted to an information center over a satellite communication system. This enables local physicians to send photographs of medical records and pictures of wounds back to Australia for a quick diagnosis.

Some other wireless monitoring systems [30], [31] have similar architectures and functions for cardiac monitoring. The general requirements and analysis of wireless patient monitoring using WLANs are presented in [8]. This includes the use of WLANs for patient monitoring in several different scenarios, requirements analysis, and design of architectures. Also, Zhou *et al.* [32] focused on the network communication techniques used by a remote surveillance platform for real-time reliable cardiac monitoring.

VII. Conclusion

Pairing a healthcare application with a wireless transport requires a thorough understanding of both the applications and the detailed functions and capabilities of the wireless technology in the context of the environment in which it will be deployed. We have proposed an architecture for enhancing the QoS of wireless ECG transmission. The basis of our approach is to split the MAC layer into MAC and LLC layers. The new MAC layer uses the IEEE 802.11 PCF mode to achieve deterministic packet delivery, and the LLC layer uses RS-based error control with block interleaving to achieve high reliability. By means of simulations using data from the MIT–BIH database, we have shown how the proposed architecture can improve wireless network performance to the extent necessary to support a telecardiology application.

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