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Real-Time Signal Processing for Fetal Heart Rate Monitoring

Muhammad I. Ibrahimy, Firoz Ahmed, M. A. Mohd Ali, and Edmond Zahedi*

***Abstract*—An algorithm based on digital filtering, adaptive thresholding, statistical properties in the time domain, and differencing of local maxima and minima has been developed for the simultaneous measurement of the fetal and maternal heart rates from the maternal abdominal electrocardiogram during pregnancy and labor for ambulatory monitoring. A microcontroller-based system has been used to implement the algorithm in real-time. A Doppler ultrasound fetal monitor was used for statistical comparison on five volunteers with low risk pregnancies, between 35 and 40 weeks of gestation. Results showed an average percent root mean square difference of 5.32% and linear correlation coefficient from 0.84 to 0.93. The fetal heart rate curves remained inside a $\pm$5-beats-per-minute limit relative to the reference ultrasound method for 84.1% of the time.**

***Index Terms*—Abdominal electrocardiogram, digital filtering, Doppler ultrasound, fetal heart rate.**

I. Introduction

FETAL heart rate (FHR) variations observed over 20 min during pregnancy and labor have commonly been used as indirect indications of the fetal condition [1]. The ability to perform long-term (e.g., 24 h) monitoring of the FHR would, thus, provide more information on the fetal condition [2].

Attempts to produce a portable system suited for long-term monitoring using Doppler ultrasound have not been successful because of its sensitivity to movements. Although there is no significant evidence from clinical data that short-term exposure to low-power ultrasound (used for imaging) is harmful to the fetus, complete safety of long-term exposure has yet to be established. To avoid these hazards, some researchers have even investigated phonocardiography applied to FHR detection [3].

Methods utilizing the abdominal electrocardiogram (AECG) have a better prospect for long-term monitoring using signal processing techniques [4]. The main difficulties encountered in determining the FHR from the AECG signal are the interference due to the maternal electrocardiogram (MECG), electromyogram (EMG), and motion artifact. To overcome the above problems, some multiple-lead algorithms use the thoracic MECG to cancel the abdominal MECG [5], but this is inconvenient for the patient during long-term monitoring.

In the present study, we have developed a system utilizing the AECG to determine the fetal heart rate. The objectives of our study are:

1) to utilize a single abdominal lead signal to extract the FHR;
2) to implement the algorithm in real-time using a small sized microprocessor-based system, making it suitable for ambulatory long-term monitoring;
3) to assess the accuracy of the developed system compared with data derived from Doppler ultrasound.

II. Methods

A. Algorithm

The algorithm was developed through a combination and modification of earlier techniques [6]–[8]. It is based on digital filtering, adaptive thresholding, statistical properties in the time domain and differencing of local maxima and minima. As shown in Fig. 1, two almost similar sets of operations are used to enhance and detect the maternal and fetal QRS complexes respectively. The AECG is first passed through a finite impulse response bandpass filter (cut-off frequencies of 10 and 40 Hz) using a Hamming window. The digital filter's coefficients have been chosen to effectively pass the highest power density of the maternal and fetal R waves. The filtered signal was then cross-correlated with an 80-ms averaged maternal QRS template, based on the normal width of the maternal QRS complex. An initial template resembling the QRS complex was first used, then continuously updated based on a running average of detected QRS complexes.

The next routine (local maxima search) records the three largest local maxima within an R wave search interval. One of the maxima is accepted as the R wave peak by the use of a thresholding technique [7] and comparison to the QRS template. These steps, performed within the validation routine, allow to discriminate between the R peak and noise. The length of the search interval is initially 1 s and it is then continuously updated after the first RR interval measurement. The threshold is updated to varying R peak and noise levels [7].

Upon detection of the maternal QRS, the corresponding MECG complex is ensemble averaged over three samples to reduce the maternal contribution from the abdominal signal. The signal is then passed through another bandpass filter (cut-off frequencies of 30 and 40 Hz) to enhance the fetal QRS complexes. The filter parameters have been selected for the steepest possible cutoffs with the requirement of real-time implementation. The differencing of local maxima and

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M. I. Ibrahimy was with the Universiti Kebangsaan Malaysia, 43600 Bangi, Selangor Darul Ehsan, Malaysia. He is now with the Department of Electrical and Electronic Engineering, Mie University, Tsu, Mie 514-8507, Japan.

F. Ahmed was with the Universiti Kebangsaan Malaysia, 43600 Bangi, Selangor Darul Ehsan, Malaysia. He is now with the Global Software Group, Motorola Australia Pty Limited, Technology Park Mawson Lakes SA 5095, Australia.

M. A. Mohd Ali is with the Department of Electrical, Electronic and Systems Engineering, Universiti Kebangsaan Malaysia, 43600 Bangi, Selangor Darul Ehsan, Malaysia.

*E. Zahedi is with the Department of Electrical, Electronic and Systems Engineering, Faculty of Engineering, Universiti Kebangsaan Malaysia, 43600 Bangi, Selangor Darul Ehsan, Malaysia (e-mail: ezahedi@vlsi.eng.ukm.my).

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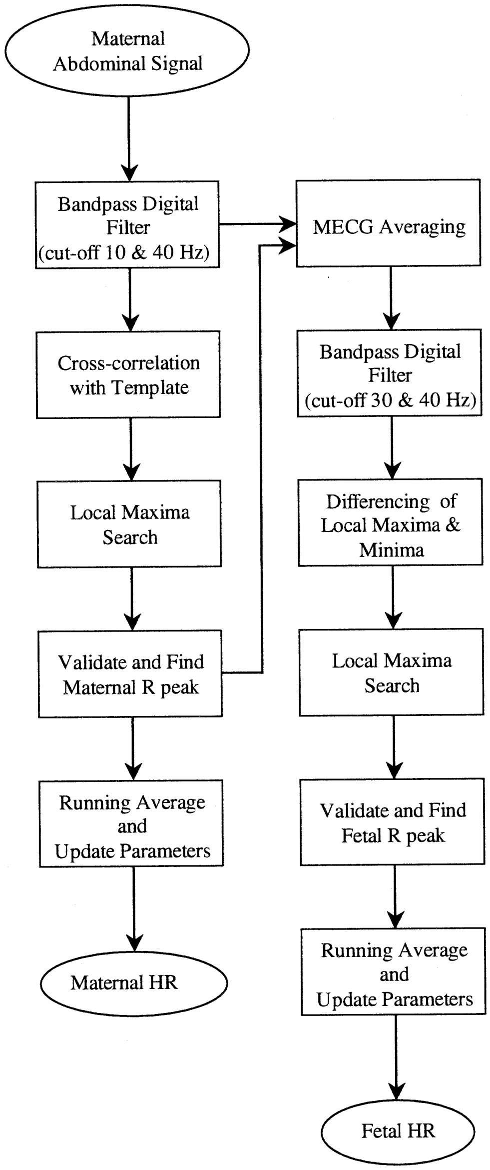

Fig. 1. General structure of the algorithm.

minima [8] is then performed on the output of the filter. A rapid and large deflection between a local maximum and the following local minimum indicates that a fetal beat has occurred. The local maxima search and validation routines are then performed similar to the maternal case. The fetal R wave search interval is first set at 640 ms assuming that initially the FHR does not exceed 187 beats per minute (BPM). The validation routine also checks for possible concurrence of the detected maximum with that of the maternal QRS complex.

B. Hardware Implementation

The algorithm was implemented on a low-power 8-bit microcontroller (PIC17C44, Microchip Technology Inc., Mountain View, CA) board. This particular microcontroller has been selected based on its features particularly well suited for this project: 8 Kbytes of EPROM, 8-bit hardware multiplication (16-bit result), addition, subtraction, shift, and logical operation in a single cycle instruction (120 ns for a 33-MHz clock input). The circuit consisting of SRAMs (128 K $\times$ 16 bits), analog-to-digital converter, amplifier (Burr-Brown INA102), power supply, and RS232 interface is shown in Fig. 2.

Single-lead maternal abdominal ECG signals were obtained using a differential amplifier by selecting the Record switch. The sampling frequency was 500 Hz and the resolution 13-bit. In the stand-alone operational mode, the measured FHR and MHR data were stored in the SRAM. Using an RS232 interface, the extracted FHR and MHR were downloaded to a PC by selecting the Transmit switch.

C. Comparison With Doppler Ultrasound

In an effort to assess the reliability of the algorithm, a set-up was designed to compare the detected FHR obtained by our system with the FHR given by a 2-MHz ultrasound fetal monitor (IFM-500, BiOSYS Co., Ltd., Seoul, Korea). A laptop computer in conjunction with a software (ICM-1000, BiOSYS Co., Ltd.) was used for logging the heart rate data (from the ultrasound fetal monitor). Simultaneous RR intervals obtained from the AECG were also downloaded from our system to the laptop. Five volunteer pregnant women with low risk pregnancies at gestational ages between 35 and 40 weeks, were studied at a clinic. All recordings were performed with the vertical electrode placement (the positive electrode on the epigastrium and the negative on the infra-umbilical), the third electrode being located on the right wrist.

A valid FHR data is obtained when a new FHR data becomes available from the ultrasound reference while the corresponding R peak is detected by our system.

III. Results

A. Hardware

The circuit boards are housed within a case (190 $\times$ 130 $\times$ 68 mm) with anodized aluminum front panel and a built-in 4 $\times$ AA battery compartment. The power consumption of the system is less than 250 mW and the net weight (including batteries) is less than 400 g. The PIC17C44 has enough processing capacity to execute the algorithm on samples taken up-to 1000 sample/sec, while achieving real-time operation. The current sampling rate is 500 Hz.

B. Comparison With Doppler Ultrasound

FHR and MHR sample traces recorded by our system are displayed in Fig. 3. When the heart rate could not be detected due to either much EMG noise during contraction, motion artifact or due to an episode of loose electrode contact, traces were left blank. The relative position of simultaneous FHR traces recorded by the Doppler ultrasound fetal monitor and FHR is shown in Fig. 4 together with their $\pm$5 BPM [3] outside limits intervals. Statistical analysis for 4164 heartbeats (mean gestation age of 37.6 weeks) using the AECG from five subjects is shown in Table I. The minimum and maximum percent root mean square difference (PRD) are 3.31% and 6.63% respectively, with a weighted average of 5.32%. The FHR curve (valid heart beats as defined in the Methods section, part C using the proposed system) remains inside the $\pm$5 BPM tolerance band on average (for all five subjects) 84.1% of the time. The minimum and maximum correlation coefficients are 0.84 and 0.93 respectively, with a weighted average of 0.89.

IV. Discussion

A weighted average PRD of 5.32% seems acceptable for this application and to the knowledge of the authors such a comparative study (with Doppler ultrasound) has not been carried-out before. The importance of the electrode–skin contact should be emphasized here as the quality of the acquired signal is of utmost importance in increasing the detection performance.

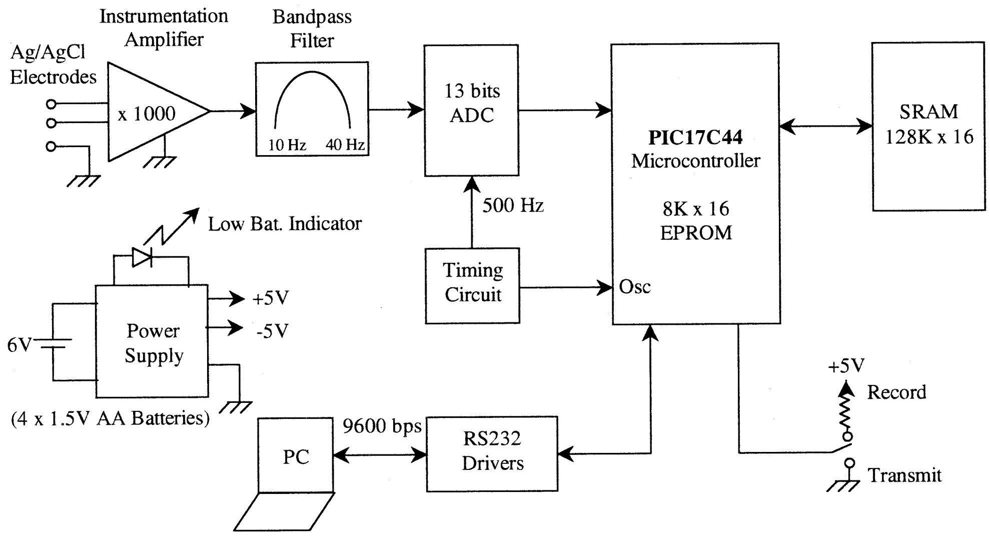

Fig. 2. Conceptual diagram of the real-time hardware.

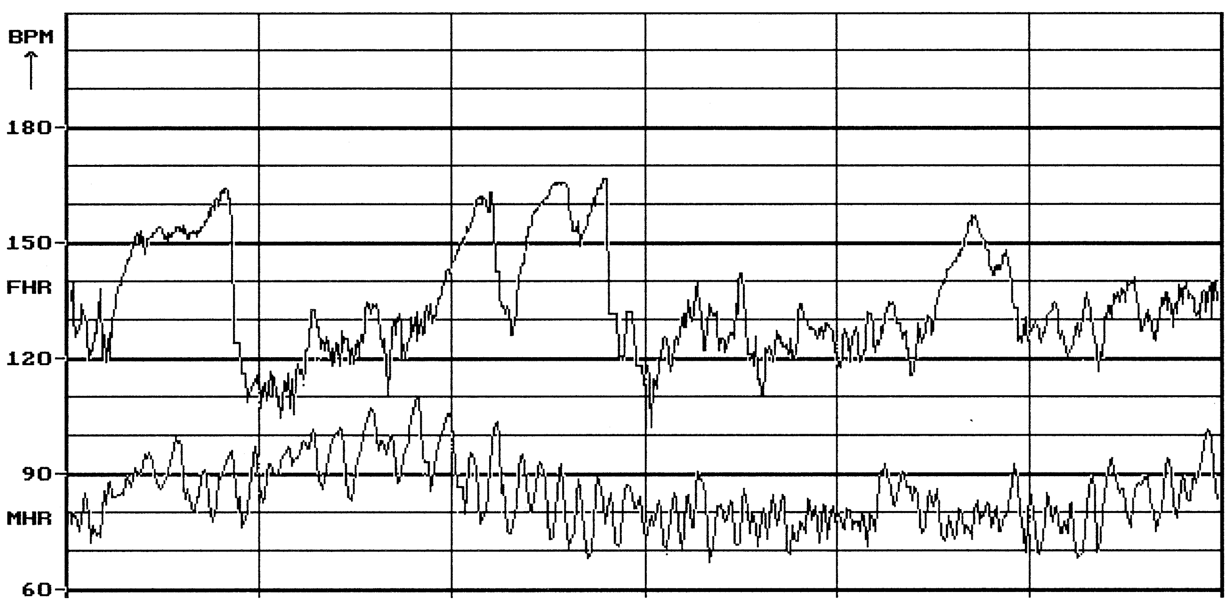

Fig. 3. Typical FHR (upper) and MHR (lower) using the proposed system (each vertical division = 90 s).

In cases where the signals are affected by noise and motion artifacts due to movements, errors are encountered in the FHR determination. Correct FHR detection is resumed when movements no longer affect the signals.

V. Conclusion

The developed system can extract both maternal and fetal heart rates in real-time utilizing a single-lead configuration. The single-lead feature is desirable from the comfort point of view of the patient especially during long-term monitoring. Although less accurate than Doppler ultrasound, our system allows for FHR as well as MHR determination in real-time, without exposing the fetus to ultrasound energy. Being of small size, weight, and power, this system is a practical alternative to ultrasound in long-term monitoring.

Some improvements to the R peak detection capability of the algorithm would be expected with enhanced procedures such as the normalization of the cross-correlation outputs. The sensitivity of the algorithm to motion artifacts and muscle noise may also be reduced with the incorporation of more rules in its RR interval validation schemes. This improvement will be a definitive step toward a fully-ambulatory system.

A favorable FECGs signal-to-noise ratio (visually detectable FECG) will allow a reliable FHR determination at a lower gestational age (below 34 weeks). An interesting direction for future research would be to compare the outcome of clinical diagnosis based on FHR/MHR determination using our system to FHR determination by ultrasound. Another area for further investigation could be the case of twins' resolution by AECG. When fully developed, such a system will be a useful tool in the assessment of the fetal condition and its relationship to that of the mother's.

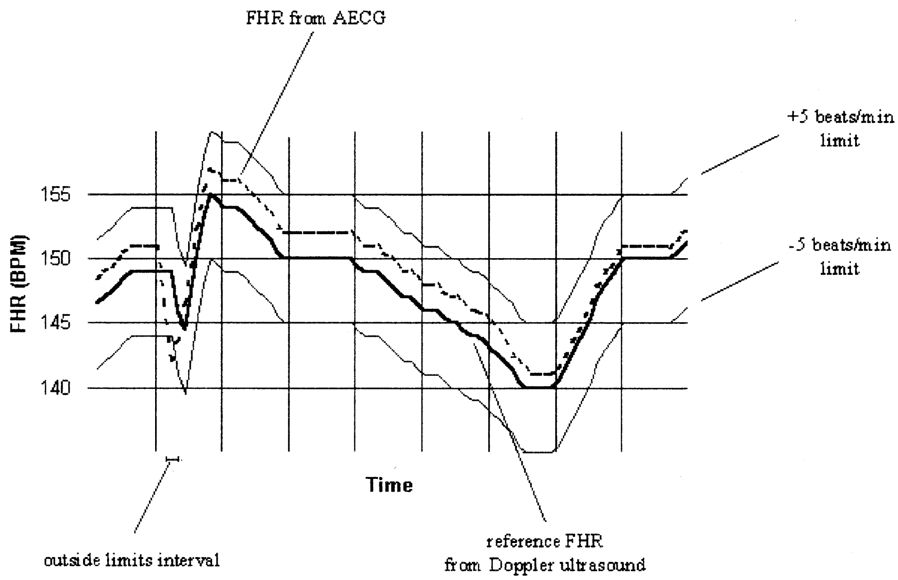

Fig. 4. Typical FHR from ultrasound (solid line) and the proposed system (dotted line) (each vertical division approximately 5 s).

TABLE I
STATISTICAL ANALYSIS OF THE SIMULTANEOUS FHR MEASURED FROM DOPPLER ULTRASOUND AND THE PROPOSED SYSTEM

Subject	Week of Gestation	No. of Valid Heartbeats	Fetal Heart Rate (BPM) MEAN ± SD & RANGE		PRD (%)	Bounded Values within (% of Total Time)		LINEAR REGRESSION
			Doppler Ultrasound	Proposed System		(±3 beats/min)	(±5 beats/min)	Correlation Coefficient R
M35	35	862	151.4 ± 6.4 (140-169)	152.7 ± 7.9 (140-202)	6.63	71.2	77.6	0.84
N36	36	995	148.7 ± 5.5 (136-168)	145.4 ± 6.9 (126-169)	3.82	86.1	91.5	0.93
O38	38	840	145.7 ± 7.3 (132-166)	147.9 ± 6.8 (122-161)	5.81	76.1	80.6	0.86
P39	39	550	136.2 ± 3.2 (130-147)	134.3 ± 4.6 (125-147)	3.31	87.8	91.8	0.92
Q40	40	917	141.0 ± 6.4 (119-162)	143.6 ± 7.7 (126-187)	6.48	77.3	80.6	0.89
All Five Subjects	37.6 ± 2.1	4164	145.3 ± 7.89 (119-169)	145.5 ± 8.8 (122-202)	5.32	79.3	84.1	0.89 0.88*

*correlation coefficient R for all cases consolidated

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Fatigue Analysis of the Surface EMG Signal in Isometric Constant Force Contractions Using the Averaged Instantaneous Frequency

A. Georgakis*, L. K. Stergioulas, and G. Giakas

***Abstract*—The averaged instantaneous frequency (AIF) is proposed as an alternative method for the frequency analysis of surface electromyography (EMG) in the study of muscle fatigue during sustained, isometric muscle contractions. Results from performance analysis using experimental EMG signals demonstrate the low variability of the proposed frequency variable. Indeed, the AIF measure is shown to perform significantly better than the widely used mean and median frequency variables, in terms of robustness to the length of the analysis window.**

***Index Terms*—Averaged instantaneous frequency, isometric contractions, mean frequency, median frequency, muscle fatigue, surface electromyography.**

I. Introduction

The detection and processing of the surface electromyography (EMG) signal offers a way of studying the myoelectric features of the neuromuscular activation associated with muscle contraction [1]. During a sustained isometric constant force contraction, the electrophysiological properties of the muscle undergo changes that are reflected in the EMG signal [1]–[3]. This phenomenon is referred to as muscle fatigue and its systematic study can provide useful information for noninvasive characterization of the behavior of individual muscles, for subjects of different gender, age, and health status. It can also provide insight into the physiology of the muscle under investigation, as well as into the mechanisms of fatigue [2], [4], [5]. The myoelectric manifestations of muscle fatigue can be quantified by monitoring the time course of characteristic variables of the surface EMG signal. These variables should facilitate efficient monitoring of the signal's frequency descent and ideally relate to the physiological processes that control fatigue.

The most commonly used spectral variables are the mean frequency (MNF) and the median frequency (MDF). Since EMG is a random signal, its power spectral density (PSD) can be estimated using classical methods (periodogram or Blackman–Tukey estimators) or modern parametric model methods [autoregressive (AR), moving average (MA), autoregressive moving average (ARMA)]. The need for a PSD estimation unavoidably introduces a number of factors (method for PSD estimation, implementation algorithm, order of parametric model, shape and size of the analysis window), which directly affect the estimates of the spectral variables [6]–[10]. Furthermore, any PSD estimation process requires a signal of time-invariant (stationary) frequency content within the analysis window. Since the EMG is a nonstationary signal [especially for contraction levels higher than 50% of maximum voluntary contraction (MVC)], one has to assume that it remains nearly stationary over short time intervals. This quasistationarity assumption introduces one more source of error to the estimation of the spectral variables.

In this paper, we propose a new frequency variable for monitoring the frequency decrement of surface EMG signals during sustained isometric constant force muscle contractions. The new EMG variable is based on the concept of the instantaneous frequency and overcomes the difficulties of the conventional spectral variables by avoiding the problem of spectral estimation. In addition, it does not require any quasistationarity assumptions, since it is inherently suitable for nonstationary signals. The proposed variable is compared with the widely used median and mean frequencies in terms of their dependence on the size of the analysis window (epoch), using experimental EMG signals.

II. Conventional Frequency Variables

The most commonly used frequency variables in EMG studies are the mean and median frequencies. The MNF is the average frequency of the power spectrum and is defined as its first-order moment

$$\text{MNF} = \frac{\int_0^\infty \omega P(\omega)\, d\omega}{\int_0^\infty P(\omega)\, d\omega} \tag{1}$$

where $P(\omega)$ is the PSD of the EMG signal and ω is the frequency variable. The MDF is the frequency at which the spectrum is divided into two parts of equal power. It can be mathematically described by the following equation, which involves zero-order moments of the PSD:

$$\int_0^{\text{MDF}} P(\omega)\, d\omega = \int_{\text{MDF}}^\infty P(\omega)\, d\omega = \tfrac{1}{2} \int_0^\infty P(\omega)\, d\omega. \tag{2}$$

These two variables have been extensively used to provide basic information about how the power spectrum changes with time.

In both methods, the EMG signal recorded during a sustained contraction is segmented into consecutive short time-windows and then PSD estimation takes place, followed by computation of the spectral variable (MNF or MDF). Thus, the time course of the MDF or the MFN is obtained. From this, the initial value and the fall rate of these spectral variables during sustained contraction are usually calculated, since they are of physiological importance [2], [4]. These parameters are estimated by fitting a least-square regression line to the MDF or MNF data points. Both linear and curvilinear regression can be used, and the parameters serve as fatigue indices (e.g., the intercept and slope of a linear regression).

III. Averaged Instantaneous Frequency

As an alternative to the conventional MDF and MNF frequency variables, we propose the averaged instantaneous frequency (AIF) measure defined as follows:

$$\text{AIF} = \frac{1}{t_b - t_a} \int_{t_a}^{t_b} \omega_i(t)\, dt \tag{3}$$

where $\omega_i(t)$ is the instantaneous frequency of the signal and the time window of calculation is $[t_a, t_b]$. The instantaneous frequency is defined as the first derivative of the signal's phase

$$\omega_i(t) = \phi'(t). \tag{4}$$

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*A. Georgakis is with the Department of Communication Systems, Lancaster University, Lancaster LA1 4YR, U.K. (e-mail: a.georgakis@lancaster.ac.uk).

L. K. Stergioulas is with the Department of Communication Systems, Lancaster University, Lancaster LA1 4YR, U.K.

G. Giakas is with the Department of Orthopaedic Surgery, University of Ioannina, P.O. Box 1330, Ioannina 45 110, Greece.

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