31st Annual International Conference of the IEEE EMBS
Minneapolis, Minnesota, USA, September 2-6, 2009

Ultra-Low-Power Wearable Biopotential Sensor Nodes

R.F. Yazicioglu, *Member,IEEE,* T. Torfs, J. Penders, I. Romero, H. Kim, P. Merken, B. Gyselinckx, H.J. Yoo, *Fellow, IEEE* and C. Van Hoof, *Member, IEEE*

***Abstract*—This paper discusses ultra-low-power wireless sensor nodes intended for wearable biopotential monitoring. Specific attention is given to mixed-signal design approaches and their impact on the overall system power dissipation. Examples of trade-offs in power dissipation between analog front-ends and digital signal processing are also given. It is shown how signal filtering can further reduce the internal power consumption of a node. Such power saving approaches are indispensable as real-life tests of custom wireless ECG patches reveal the need for artifact detection and correction. The power consumption of such additional features has to come from power savings elsewhere in the system as the overall power budget cannot increase.**

I. Introduction

WIRELESS sensor nodes and networks are spanning an ever increasing range of applications. They are now used in industrial applications such as process monitoring and process control, machine health monitoring, but also in civilian applications such as environmental monitoring, traffic control, home automation, and they are emerging in personal healthcare applications with a focus on prevention rather than cure [1]. This paper focuses on Body Area Network (BAN) [2] nodes that are intended for personal health monitoring and assisted living, but also branch into lifestyle, sports and entertainment applications.

Key challenges of every sensor node are power and functionality. The first challenge could be summarized as 'ultra-low-power (ULP) everywhere' and stems from the vision that these systems should have energy autonomy. As such, the power challenge impacts on the complete system design of the node ranging from the sensor and its front-end to the signal processing to the wireless communication. The second challenge relates to biological and physiological sensing, which needs to be achieved equally with ULP dissipation [3]. This challenge is being met by biopotential sensors and also by physical sensors such as accelerometers.

This paper focuses on personal healthcare monitoring nodes using biopotential interfaces [4]. Several generations are summarized and a continuous reduction in power consumption is seen. Power savings come from the analog sensor front-ends by architectural improvements and from the localized signal filtering and data compression. Evaluations of our wireless ECG patch show that depending on activity, there is a need for artifact detection and removal.

C. Van Hoof, J. Penders, I. Romero and B. Gyselinckx are with the HOLST Centre, High Tech Campus 31, 5656 AE Eindhoven, The Netherlands. Corresponding author is Chris Van Hoof (phone: +3216281815, fax: +3216288500, e-mail: chris.vanhoof@imec.be).

R.F. Yazicioglu, T. Torfs, P. Merken and C. Van Hoof are with IMEC, Kapeldreef 75, 3001 Leuven, Belgium.

H. Kim and H.J. Yoo are with the Department of Electrical Engineering and Computer Science, KAIST, Daejeon, 305-701, Republic of Korea.

II. Sensor platform

Our wireless sensor platform consists of IMEC's proprietary designs as well as commercial off-the-shelf components. As commercial components it uses a Nordic nRF24L01 radio chip-set and a TI MSP430f1611 micro-controller as well as a 160mAh Li-ion battery. The customized components are a biopotential signal amplifier circuit, an antenna compatible with on-the-body sensing, and a power management circuitry to optimize the power performances of the sensor node.

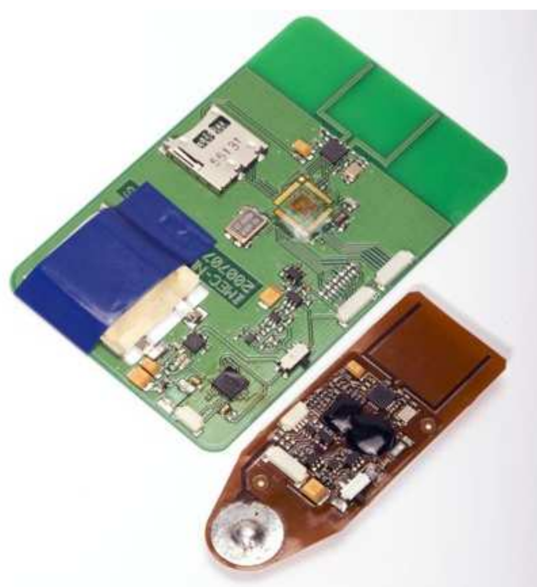

FIG. 1. Wireless ECG patch integrated on flexible substrate (bottom). Depending on the application, the receiver is a wireless USB stick (for continuous data streaming to a base station, not shown) or a receiver card with built-in flash memory (for data logging, shown on top).

Several embodiments have been made: a cube of approximately 1cm^3 [5][6][7] which can be used as a generic platform (integrating other sensors) as well as a thin flexible patch [8] mainly intended for biopotential ExG (ECG, EMG, EEG) recordings [9][10]. The flexible form factor was particularly suitable as a wireless flexible ECG patch, as illustrated on *FIG. 1*. Power consumption of the wireless ECG system, while transmitting raw ECG signal to a receiver within 10m range, is 1.17mW in data streaming mode at 3kbps. This level of power consumption has been obtained thanks to the ULP front-ends presented further in this article. Depending on the application, the power consumption can or cannot be further reduced by duty-cycling. Although these power levels are state-of-the-art for wireless biopotential monitoring, lower overall power consumption in the order of 100uW is needed for a more compact system allowing a smaller battery. This implies the need for a lower-power (i.e. dedicated) signal processing solution and a lower-power radio solution.

III. ULTRA-LOW POWER BIOPOTENTIAL FRONT-ENDS

The sensing and read-out of the biomedical signals initially drew a significant part of the power budget. The acquisition of biopotential signals is specifically complicated by a (varying) differential electrode-offset and by 50/60Hz interference. This necessitates AC-coupled low-noise amplifiers with high common-mode rejection ratio. Our first generation front-end was a current-balancing instrumentation amplifier (CBIA) [11] which consumed 0.8mW for a two-channel EEG (or ECG) recording. Supply voltage of our biopotential amplifiers is 3V.

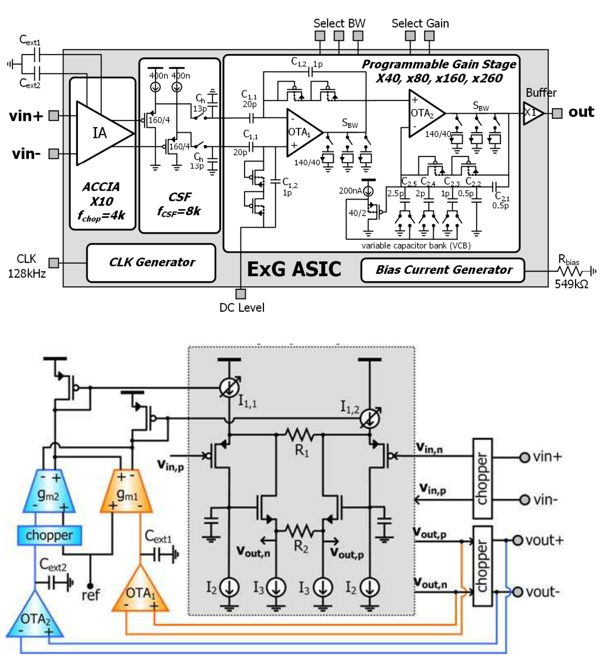

FIG.2. Overall 2nd generation front-end (top) and simplified architecture of the core ACCIA (bottom).

An improved 2nd generation solution was found in an AC-coupled chopper stabilized instrumentation amplifiers (ACCIA). This ACCIA can achieve high-CMRR and eliminate 1/f noise while filtering the differential electrode offset by using a DC servo-loop outside the choppers that introduces a high-pass filter (HPF) characteristic [12]. The power consumption of such 2-channel front-end amounted to only 0.12mW. The block-diagram of the overall 2nd generation architecture and the simplified ACCIA architecture are shown in *FIG. 2*. An optimum 3rd generation further reduced the power level of the front-end to 0.025mW by replacing the DC current servo-loop by a fine analog and coarse digital servo [13]. This generation also featured a 12bit ADC whose power consumption is included. *FIG. 3* shows the power consumption breakdown between front-end, microcontroller and radio for the three generations. It is obvious that the analog front-end power consumption has dropped from 30% of the overall system power to 3% of the system power. The microcontroller power consumption has slightly decreased between generation 2 and 3 due to the built-in ADC in the 3rd generation front-end and the wireless has remained unchanged. In view of the above, future efforts should focus on reducing the power consumption of the microcontroller (i.e. signal processing and radio).

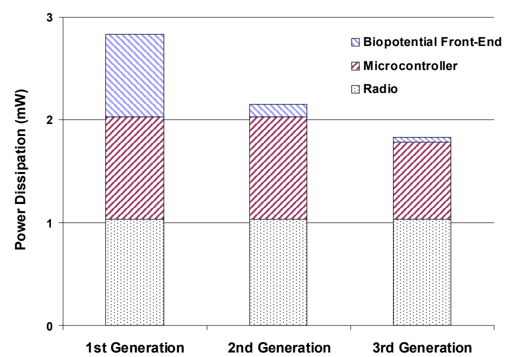

FIG.3. Power consumption of three generations of 2- channel ExG wireless sensor nodes, for 340Hz sampling rate per channel.

IV. POWER CONSUMPTION TRADE-OFFS

The data shown in *FIG. 3* were achieved with a 340Hz sampling rate per channel. As the radio power consumption scales proportionally with sampling rate, reducing the sampling rate is an obvious way to further reduce the overall system power, provided signal integrity remains. In our current sensor nodes, raw data is transmitted and this causes the wireless communication to take up 50% of the power budget. For the specific case of ECG delineation, we have compared data processing performed on the local node and on the base station. A complete real-time delineation of the ECG wave-forms with improved T and P wave peak, onset, and end search is done [14]. The algorithm extracts the onset, peak and off-set of the P, QRS and T waves out of the ECG signal. As ECG signal source we use the MIT QT database. It was expected that the reduction in data transmission would lead to power savings. However, the implementation on the sensor node platform with nRF24L01 radio chip-set and MSP430f1611 microcontroller does not achieve power savings, as shown in *TABLE 1*. The increase in local computation counteracts the limited savings in the radio which at these low data rates and duty cycles (<0.1%)

is dominated by idle power consumption. To reduce the node power consumption, we target the development of dedicated and power optimized signal processing platforms.

TABLE 1
DATA STREAMING VS LOCAL ECG PROCESSING

Power consumption	ECG data streaming (3kbps)	Local processing and streaming at 100bps
Microcontroller	0.39mW	1.12mW
Radio	0.77mW	0.62mW
Radio Transmit	0.16mW	0.01mW
Radio Wake-up	0.02mW	0.02mW
Radio Idle	0.59mW	0.59mW
Total	1.16mW	1.74mW

In [15], we reported the design of a dedicated DSP for ULP applications, achieving the implementation of an R-peak detection algorithm consuming as little as 8.4 µW, for a 200Hz sampling frequency and 16-bit resolution. To further increase the functionality we will incorporate a dedicated ECG signal processing circuit for arrhythmia monitoring (*FIG. 4*). This circuit has been implemented in a 0.18µm UMC CMOS process and is designed to dissipate 4µW per 1kHz sampling frequency and 12-bit resolution [16].

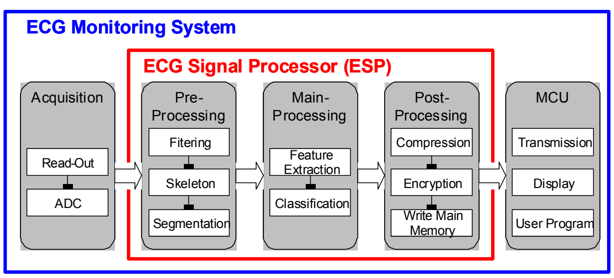

FIG.4. Concept of ECG monitoring system including local ECG processing using an arrhythmia monitoring ASIC.

FIG. 4 shows the concept of an ECG monitoring system incorporating the ECG signal processor. The local processing not only includes the quadratic level ECG compression algorithm (QLC) presented in [17] but also the capability of R peak detection, arrhythmia monitoring, and data encryption. Hence, the combination of arrhythmia detection with signal compression can lead to significant reduction in data transmission over the wireless link. For instance, in one scenario, only the sections of the ECG signal with abnormal ECG signal can be transmitted, where the ECG waveform is already compressed using the QLC.

Lowering the radio consumption is also essential for a further reduction in power consumption. Compared to narrow-band transceivers, Ultra-Wide Band (UWB) transceivers can provide significant power savings. The FCC has recently authorized UWB communications between 3.1GHz and 10.6GHz and the emerging 805.15.4a standard has subdivided the entire UWB spectrum in 500MHz sub-bands. Low-power and low-complexity implementations of a transmitter complying with the 802.15.4a specifications are proposed in [18][19], which translate to a power efficiency of roughly 1nJ/bit, which compares favorably to the 20nJ/bit for the nRF24L01. In [20], a low complexity receiver was reported with an energy efficiency of approximately 10nJ/bit, which compares to narrowband receivers. Such asymmetric power consumption is adequate for BAN transceiver applications since the most power-constrained elements are the nodes and not the base station.

V. SYSTEM TESTS

The wireless ECG patches have been tested in ambulatory settings, to evaluate the influence of physical activity on the quality of ECG recordings.

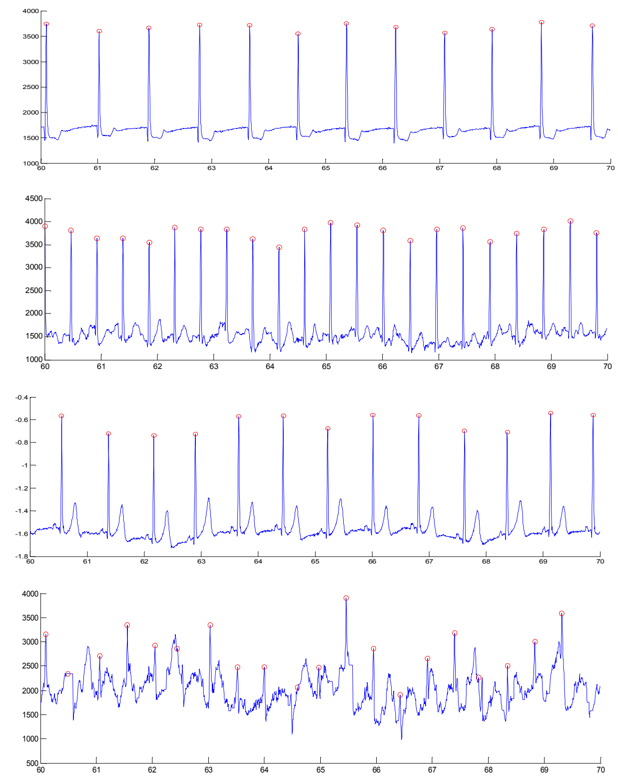

FIG. 5. Ambulatory ECG recording samples. The two top graphs correspond to the ECG patch integrated in a chest belt for a low and a high level of physical activity (top vs bottom). The two bottom graphs correspond to the ECG patch integrated in a textile pocket in similar conditions.

10 healthy volunteers have been involved in the study. Three levels of physical activity are considered: at rest (low level of activity), biking on a static bike at 100 W and 70 RPM (moderate level of activity), and running on a treadmill at 7.5 km/h (high level of activity). The ECG patch was used in two embodiments: a textile pocket containing the ECG patch and attached to the chest via standard Ag/AgCl electrodes and a chest belt integrating the ECG patch, using electrode positions corresponding to standard leads II and I respectively. Recordings of 10 minutes are obtained for each level of activity, for each packaging and for each patient. FIG. 5 shows traces of those recordings. The SNR can be considered to vary from +10dB for a low-level activity to 0dB for moderate activity, and down to -10dB for high level

of activity with the textile pocket. The stronger deterioration in case of a more movable embodiment may imply that relative motion between patch and body is causing larger artifacts than absolute motion. The reduction in signal-to-noise ratio has to be prevented as it will hamper feature extractions such as QRS detection and rhythm analysis [21]. Motion detection is one approach to determine motion-related artifacts in ECG recordings, as shown in FIG. 6. It shows the simultaneous recording of accelerometers and ECG signal which allows removal of motion-related artifacts. However, accelerometer-based motion detection is able to detect absolute motion and not relative motion. For this reason, alternative methods should be pursued.

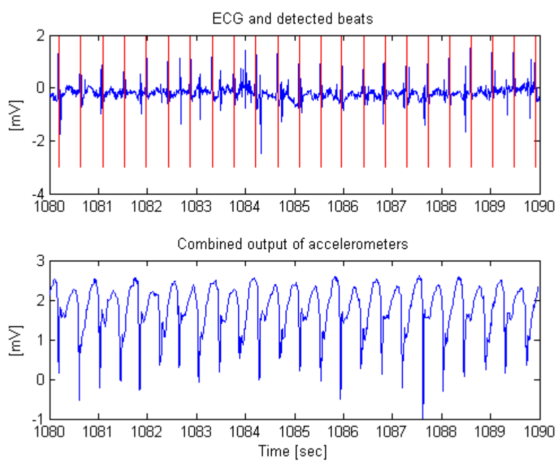

FIG. 6. Combined output of multi-axis accelerometers and ECG allows classifying motion-related artifacts and removing those from beat detection.

VI. CONCLUSION

Ultra-low-power consumption challenges all building blocks of wearable sensor nodes. It was shown that the telemetry of raw biopotential signal results in a clear bottleneck in radio power dissipation. This necessitates alternative radio architectures with higher power efficiency or local feature extraction to reduce the amount of wireless data communication.

Further power reduction may be possible by actively incorporating the analog building blocks in this signal processing scenario and already minimize the data rate as much as possible before the DSP block, i.e. in analog domain or in the level of analog-to-digital conversion.

Practical use of these wireless sensor nodes will require dry biopotential electrodes. The relatively high impedance and large half-cell potential of dry electrodes needs readout circuits that are able to filter large DC offsets while sustaining high CMRR.

REFERENCES

[1] B. Gyselinckx, Ph. Mattelaer, R. Borzi, "Human++: emerging technology for Body Area Networks", in *Brave New Interfaces. Individual, Social and Economic Impact of the Next Generation Interfaces*, Crosstalks, 2007

[2] R. Schmidt, T. Norgall, J. Mörsdorf, J. Bernhard, T. von der Grün, “Body Area Network BAN, a key infrastructure element for patient-centered medical applications”, *Biomed Tech (Berl)*. 2002;47 suppl 1 pt 1:365-8

[3] Guang-Zhong Yang, *Body Sensor Networks*, Ed. Springer, 2006.

[4] J. Penders, B. Gyselinckx, R. Vullers, M. De Nil, J. van de Molengraft, P. Merken, C. Van Hoof, “Human++: from technology to emerging health monitoring concepts”, in *Proc. of the 5th Int. Workshop on Wearable and Implantable Body Sensor Networks*, China, p94-98 (2008).

[5] H. Diall, K. Raja, I. Daskalopoulos, S. Hailes, G. Roussos, T. Torfs, C. Van Hoof, “Sensor Cube: A Modular, Ultra-Compact, Power-Aware Platform for Sensor Networks”, IPSJ-DC Vol. 3, pp. 309-319, 2007.

[6] B. Gyselinckx, R. Vullers, C. Van Hoof, J. Ryckaert, R.F. Yazicioglu, P.Fiorini, V.Leonov, “Human++: Emerging Technology for Body Area Networks”, IFIP International Conference on Very Large Scale Integration, pp. 175-180 (2006).

[7] K. Baert, B. Gyselinckx, T. Torfs, V. Leonov, F. Yazicioglu, S. Brebels, S. Donnay, J. Vanfleteren, E. Beyne, C. Van Hoof, “Technologies for highly miniaturized autonomous sensor networks”, Microelectronics Journal Vol. 37 (12), pp. 1563-1568 (2006).

[8] J. Penders, R. Vullers, G. Dolmans, M. Ashouei, P. Merken, V. Leonov, B. Gyselinckx and C. Van Hoof “Human++: Technologies for Body Area Networks and Emerging Applications”, *IEEE Trans. of Inform. Technol. in Biomed.*, submitted

[9] R. F. Yazicioglu, P. Merken, R. Puers et al.: :A 60 µW 60 nV/√Hz Readout Front-End for Portable Biopotential Acquisition Systems,” *IEEE J. Solid-State Circuits*, pp. 1100 – 1110, May 2007.

[10] N. de Vicq, F. Robert, J. Penders, B. Gyselinckx, T. Torfs, “Wireless Body Area Network for Sleep Staging”, in *Proc. Int. Conf. on Biological Circuits and Systems*, 2007.

[11] R.F. Yazicioglu, P. Merken, C. Van Hoof, “Integrated low-power 24-channel EEG front-end”, Electron. Lett. Vol. 41 (8), pp. 457-458 (2005).

[12] R. F. Yazicioglu, P. Merken, R. Puers, C. Van Hoof, “A 60 µW 60 nV/√Hz Readout Front-End for Portable Biopotential Acquisition Systems,” IEEE J. Solid-State Circuits, pp. 1100 – 1110, May 2007.

[13] R. F. Yazicioglu, P. Merken, R. Puers, C. Van Hoof, “A 200µW Eight-Channel Acquisition ASIC for Ambulatory EEG Systems” ISSCC Dig. Tech. Papers, pp. 6-7, Feb. 2008

[14] S. Nimmala, J. van de Molengraft, J. Penders and B. Gyselinckx, ”An intelligent wireless ECG patch for single-lead ECG monitoring”, to be presented at the 33rd annual conference from the International Society of Computerized Electrocardiography, Riverside CA, USA, 2008.

[15] L. Ysedoobt, M. de Nil, et al., Design of ultra-low-power wireless sensor nodes for biomedical monitoring, in Proc. Of International Symposium on Systems,Architectures, Modeling and Simulation (SAMOS VII), 2007.

[16] H. Kim, R.F. Yazicioglu, T. Torfs, P. Merken, C. Van Hoof, and H.J. Yoo, “An Integrated Circuit for Ambulatory Arrhythmia Monitoring Systems,” IEEE Proc. of EMBC, (submitted).

[17] H. Kim, Y. Kim, and H.J. Yoo, “A Low Cost Quadratic Level ECG Compression Algorithm and Its Hardware Optimization for Body Sensor Network System,” IEEE Proc. of EMBC, Aug. 2008

[18] Y.H. Choi. Gated UWB pulse signal generation. In IEEE Joint Int. Workshop of UWBST and IWUWBS, pages 122–124, May 2004.

[19] J. Ryckaert, M. Badaroglu, C. Desset, V. de Heyn, G. Van der Plas, P. Wambacq, B. Van Poucke, and S. Donnay. Carrier-based UWB impulse radio: Simplicity, flexibility, and pulser implementation in 0.18µm CMOS. Int. Conference on Ultrawideband, ICU 2005, 2005.

[20] J. Ryckaert, M. Badaroglu, V. De Heyn, G. Van der Plas, P. Nuzzo, A. Baschirotto, S. D’Amico, C. Desset, H. Suys, M. Libois, B. Van Poucke, P. Wambacq, and B. Gyselinckx. A 16mA UWB 3-to-5Ghz 20Mpulses/s quadrature analog correlation receiver in 0.18µm CMOS. In ISSCC Digest of Technical Papers, pages 114–115, February 2006.

[21] I. Romero Legarreta, P. Addison, et al. “Continuous wavelet transform modulus maxima analysis of the electrocardiogram: Beat characterization and beat-to-beat measurement,” International Journal of Wavelets, Multiresolution and Information Processing, vol. 3, no. 1, pp. 19–42, 2005.