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# NONINVASIVE MULTIPARAMETER REGISTRATION SYSTEM FOR EVALUATING PHYSIOLOGICAL CONDITION

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**Abstract.** The aim of this work is to create a system for simultaneous registration of different physiological parameters to evaluate the state of health of a patient and its correlation with the parameters. Several physiological signals are registered and analysed: electrocardiogram, bioimpedance and the movements' intensity of the body. As a result of the simultaneous and noninvasive recording and analysis of these signals, a lot of physiological parameters like heart rate, QT-interval, ventilation volume and heart impedance changes can be compared. In addition to that, monomorphic action potential duration and QT-interval are compared to determine the correlation between parameters.

**Key words:** ECG, heart rate, QT-interval, bioimpedance, simultaneous measurements.

## 1. INTRODUCTION

There are many different possibilities to estimate the physical state or the health condition of a patient. For example, ECG with its frequency and different time intervals, oxygen consumption, blood pressure, chemical analysis of the blood, etc.

Simultaneous registration and analysis of a number of parameters needs different equipment and, moreover, many of the parameters require invasive registration. Signals that are noninvasively measurable are more convenient in clinical practice, even though in several cases they are less informative. Simplicity and much lower risk to harm the patient makes them beneficial.

Heart is probably the most important organ to reflect the total health condition of a patient. Cardiac functions are mainly controlled by autonomous nervous

system (ANS). It controls the heart directly by nerve pulses and also by its mediators like cathecholamines, by changing their concentration in blood. It is practically impossible to measure ANS information directly and noninvasively, but reactions of the heart to ANS are measurable in most cases and, moreover, they are measurable electrically. It means that the system of measurements is relatively simple, safe and fast.

In clinical practice, ECG is widely used as a noninvasive procedure to test the heart. As the present work is mainly focused on electrically measurable parameters, the ECG method is chosen as a basis of the investigation. Heart rate (HR) is certainly the most important parameter of ECG but there are lots of other parameters reflecting the physical state of the heart like QT-interval (QT) which also carries ANS information [1,2]. For example, endocardially recorded monomorphic action potential (MAP) is widely used in pacemakers to adapt the paced heart rate to metabolic demands of the patient [3]. Unfortunately, invasive and complex character of the registration procedure limits repeated performance of direct MAP registration. The QT, obtained from a standard ECG, should be well correlated with MAP because it represents the same phase of heart cycle as endocardially recorded MAP. The QT with its prolonging/shortening effect, variability and spatial dispersion has become very popular by evaluating the physical state of the heart [4].

Respiratory parameters are also very informative and widely used in sports medicine. Oxygen consumption is a relevant measure to characterize the subject's work state but it is not a directly electronically measurable parameter. It has been reported that ventilation rate and volume have a high correlation with oxygen consumption [5]. These parameters are measurable by detecting bioimpedance changes in the subject's torso [6]. Bioimpedance measurement gives us an alternative and a relatively simple tool to monitor ventilation parameters. Using the bioimpedance method it is also possible to detect impedance changes in myocardial impedance. These changes are relatively small compared with respiratory changes, but generally detectable.

A convenient reference device to determine the workload of the patient is piezoelectric physical activity detector (which is widely used in rate adaptive cardiac pacemakers) [3].

Simultaneous registration of different parameters can give better picture of physiological changes during exercise than using only one of them.

## 2. MATERIALS AND METHODS

A multichannel recording system has been developed for simultaneous electrical measurement of ECG, bioimpedance of thorax and myocardium, and body motions (Fig. 1). The analogue part of the recording system consists of the following.

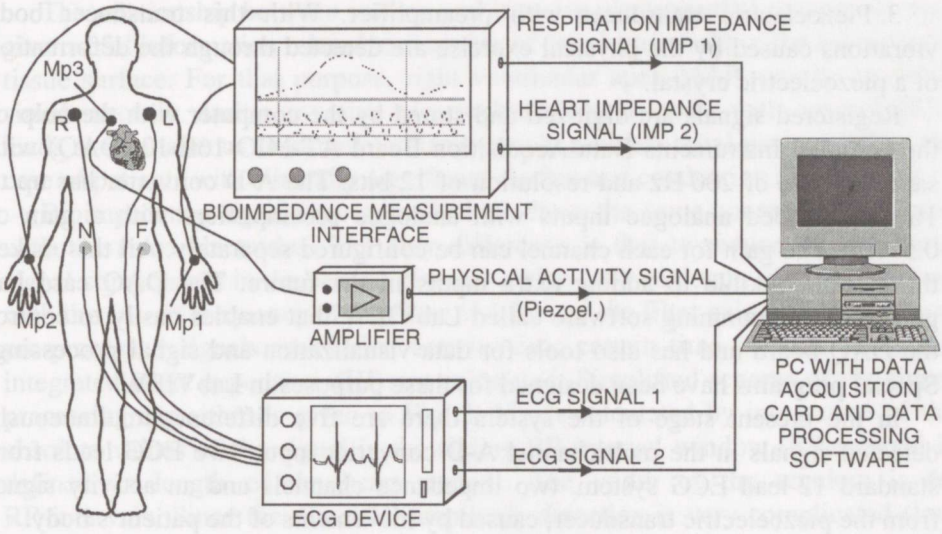


**Fig. 1.** Schematic illustration of the multichannel recording system for registration of different physiological parameters.

1. Standard 12-leads ECG recorder (Siemens); so far only two channels (with maximum T-wave amplitude) have been used for digital recording purposes but theoretically all 12 ECG leads can be stored by computer. From 12 ECG leads the leads with maximum T-wave amplitude were digitally recorded. In Fig. 1 only limb leads are shown but actually all 12 leads are connected to the patient.

2. Bioimpedance measurement interface. The interface is based on lock-in measurement method; it has current source <80 μA and frequency of excitation of the sine wave 100 kHz [7,8]. The system with two electrodes is used for impedance measurement. In our case electrodes are placed on both arms (Mp1 and Mp2 in Fig. 1). The thoracic bioimpedance varies with breathing and also with contractions of the heart. The main harmonic of the signal component which reflects breathing (breathing component) is assumed to lay in the frequency range of 0.3–1 Hz. The main harmonic of the signal component reflecting the heart beat (heart beat component) is assumed to be in the range of 1–3 Hz or 60–180 beats per minute. Values exceeding these limits are possible, but the possibility of their occurrence is small. Waveforms of bioimpedance signals can not be assumed to have a harmonic shape. Moreover, the higher harmonics of the breathing component can be in the frequency range of the main harmonic of the heartbeat component (1–3 Hz). This makes separation of the components complicated. Magnitude of the heartbeat component has a correlation with stroke volume of the heart [9]. In order to separate signal components caused by heart beating and respiration, an additional coupling is used. This coupling contains a low-pass filter with the cut-off frequency of 1 Hz.

3. Piezoelectric transducer with preamplifier. With this transducer body vibrations caused by the physical exercise are detected through the deformation of a piezoelectric crystal.

Registered signals are digitized and stored by the computer with the help of the National Instruments Data Acquisition Board AT-MIO-16E-10 (DAQ) with sampling rate of 200 Hz and resolution of 12 bits. The A-D converter has up to 16 single-ended analogue inputs with analogue preamplifiers with a gain of 0.5–100. The gain for each channel can be configured separately. All this makes the system flexible to adding extra inputs in the future. The DAQ card has graphical programming software called LabVIEW that enables easily to control the DAQ board and has also tools for data visualization and signal processing. Special programs have been designed for these purposes in LabVIEW.

In the present stage of the system there are five different simultaneously detected signals in the multichannel A-D converter input: two ECG leads from standard 12-lead ECG system, two impedance channels and an activity signal from the piezoelectric transducer, caused by movements of the patient's body.

The instantaneous HR and QT are calculated from the ECG signal. To get instantaneous changes of the parameters we have to detect every single heartbeat [10]. Because of the fact that exercise ECG can be often distorted by breathing and muscle artifacts, the baseline correction is needed before calculation of HR and QT. For baseline correction we used high-pass finite impulse response filter [11], with cut-off frequency of 0.8 Hz.

It is always problematic to determine the end of the T-wave. In this work the following T-end fixation algorithms are tested.

1. The classical definition of T-end is the point where ECG converges with isoelectric baseline. This point can not be used for reliable automatic detection since often in exercise ECGs with high HRs the T-end does not reach the isoelectric baseline at all.

2. The T-end is considered to be the point where tangent at the maximum absolute value of the derivative of the T-wave and baseline intersect. This method is better than the first one but the number of misdetections is still quite large [12].

3. The end of QT is taken at the point of the T-wave maximum (T-apex). This method gives the best result because the aforementioned point is easily detectable. Absolute values of the QT in this case are shorter than the QT measured from the T-end but this shift has practically no influence on the dynamical part of QT [13].

Before the QT is being included as a parameter in current investigation, another series of experiments is carried out. During this investigation a possible relationship between the QT duration and the duration of endocardially recorded MAP is determined to find out whether the dynamics of the QT, obtained from standard ECG lead, is comparable with changes of the right ventricular MAP duration at spontaneous as well as stimulation-induced HR variations [14].

The relationship between surface and invasive electrocardial parameters may give useful information related to mapping of the electric field of the heart on the tissue surface. For that purpose, right ventricular apex MAP may be recorded simultaneously during invasive electrophysiological study with several ECG leads at spontaneous or stimulated (transesophageal left atrial pacing) HR in the case of patients with clinically significant rhythm and conduction disturbances.

Bioimpedance signals are actually both from the same source and detected with the same electrodes. The only difference is that to reduce the breathing signal components in the second channel a high-pass analogue filter is designed with the cut-off frequency of the filter of 1 Hz and the filter gain of 5. From these pre-processed signals ventilation activity over certain time interval (VV) and integrated heart impedance (HI) are calculated. Simplified scheme of the signal processing algorithm is given in Fig. 2. VV is calculated by integration of the absolute values of the signal in the sliding RR-interval window (RR-interval is defined as length of the heart period). The width of the window is 60 RR-intervals. Since the direct HI amplitude detection is very complicated (low level of the signal and a lot of artifacts), the signal is integrated as well but in this case the integration window is 10 RR-intervals. Using integration window with the length of only one (current) RR-interval may give more precise results but in that case the results are too noisy, i.e., the trend of the signal is not sufficiently clearly visible and later averaging is necessary.

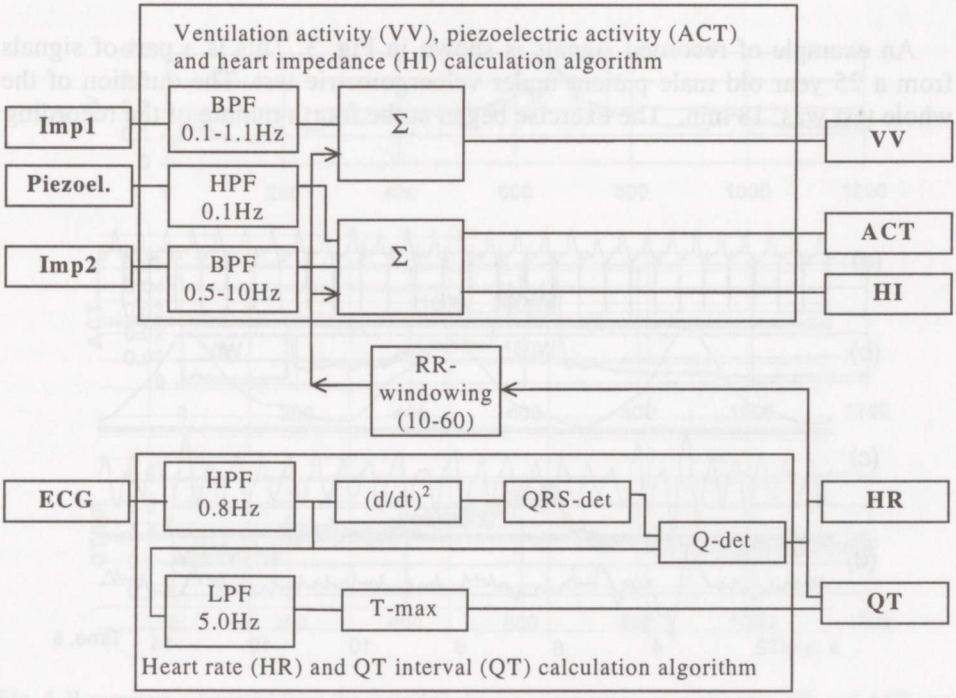


**Fig. 2.** Block diagram of the signal processing algorithm.

Piezoelectric activity signal (ACT) that is mainly stored for reference purposes to reflect work load is also integrated in the sliding window with the length of 10 RR-intervals. It is impossible to calculate the absolute workload in such a manner but relative changes are observable.

The sliding RR integration windows are used to synchronize signals with detection of HR and QT. It means that every time there is a heart beat, other signals get new values as well. This is especially useful because in this way it is possible to visualize and analyse signals simultaneously even in real time.

The duration of a recording with our experimental set-up can be as long as necessary, i.e., until there is free space on the hard disk, because all data are collected and written continuously on the hard disk in real time. The recorded file format is binary, no data reduction algorithms are used.

Different types of exercises can be used (bicycle and treadmill tests, etc.) to test the patient in different conditions. Recording software monitors only current HR of the patient and lets us know when the exercise should be finished; other parameters will be calculated after the recording is completed. Every recorded data file contains also personal data about the patient as name, patient's code, age, and also the type of the test protocol. This makes it easier to handle large databases because every test can be reconstructed.

## 3. RESULTS

An example of recorded signals is shown in Fig. 3. This is a part of signals from a 25 year old male patient under veloergometric test. The duration of the whole test was 18 min. The exercise began at the fourth minute of the recording

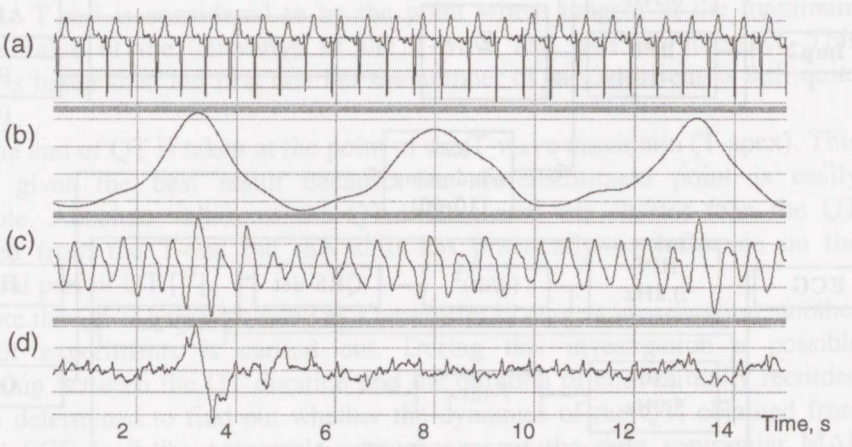


**Fig. 3.** A fragment of recorded signals: (a) ECG; (b) ventilation impedance; (c) heart impedance; (d) signal from the piezoelectric transducer.

at 50 W work load and after every three minutes the work load was increased by 50 W until the submaximum of HR was achieved, thereafter a ten minute recovery period followed.

Parameters calculated from the results of the recording are shown in Fig. 4. The exercise was started at 180 and stopped at 680 s. During the exercise the workload was increased in three steps. These steps can be seen in the ACT signal. It is interesting to mention that when the workload is high (150 W) and the patient is already tired, the physical activity increases considerably compared with lower workload.

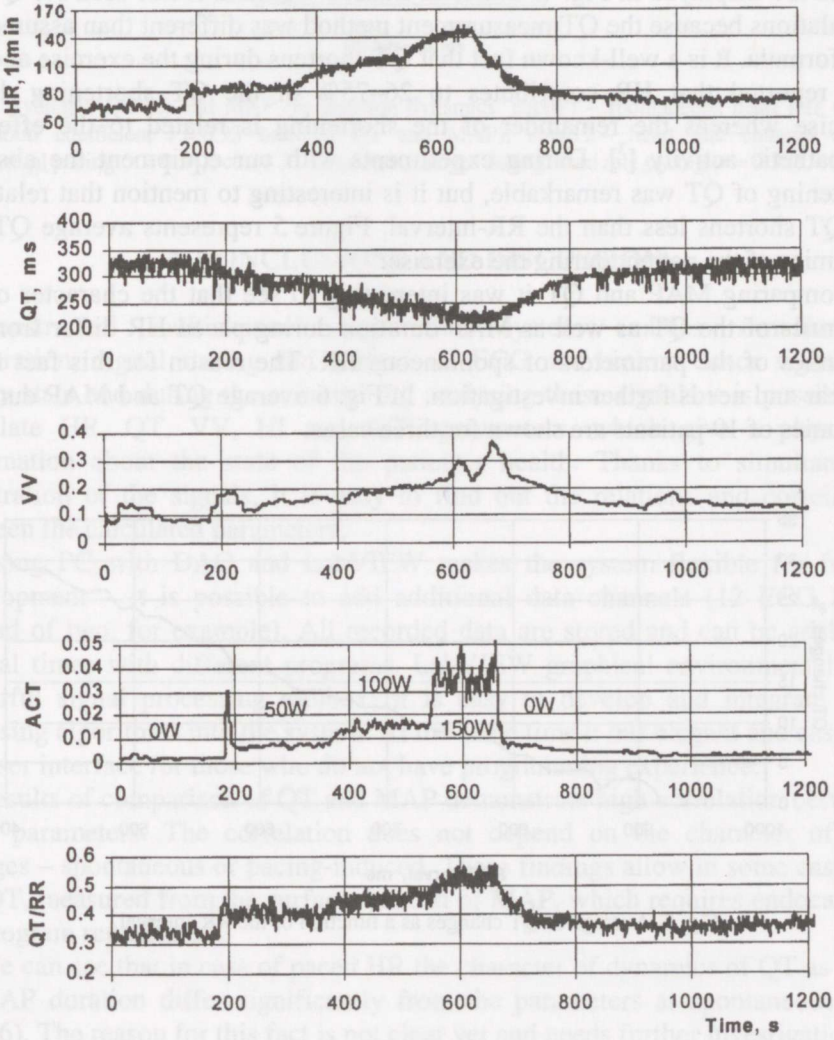


**Fig. 4.** Parameters calculated from the recorded signals of the veloergometric test: VV and ACT are shown in relative units.

The parameter VV characterizes ventilation volume changes during the physical activity. Initially the ventilation frequency was also calculated but the results showed that frequency changes are not as informative as volume changes. In most cases the ventilation frequency changes are negligible.

Despite of the fact that HI signal detected in the steady state of the patient has good quality, the HI parameter registration during the physical exercise with this experimental set-up is almost impossible. The reason for it is very small signal amplitude and there are lots of artifacts in the same frequency band caused by body movements (the bioimpedance measurement is sensitive to body motions). That is the reason why instead of the HI signal, relative QT (QT/RR) shortening values are displayed in Fig. 4. Classical Bazetts' formula is not used for QT/RR calculations because the QT measurement method was different than assumed by that formula. It is a well-known fact that QT shortens during the exercise and it is also reported that HR contributes to 26–75% of the QT shortening during exercise whereas the remainder of the shortening is related to the effect of sympathetic activity [3]. During experiments with our equipment the absolute shortening of QT was remarkable, but it is interesting to mention that relatively the QT shortens less than the RR-interval. Figure 5 represents average QT–RR dynamics of the patient during the exercise.

Comparing MAP and QT it was interesting to see that the character of the dynamics of the QT as well as MAP duration during paced HR differ from the dynamics of the parameters of spontaneous HR. The reason for this fact is yet unclear and needs further investigation. In Fig. 6 average QT and MAP duration dynamics of 19 patients are shown for three cases.

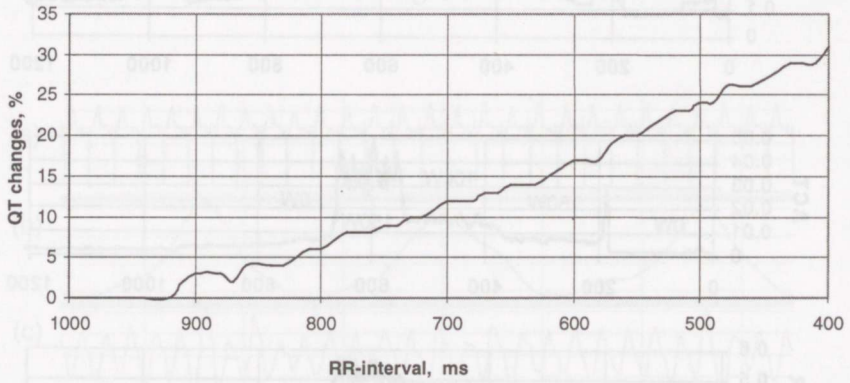


**Fig. 5.** Average of QT changes as a function of the RR-interval.

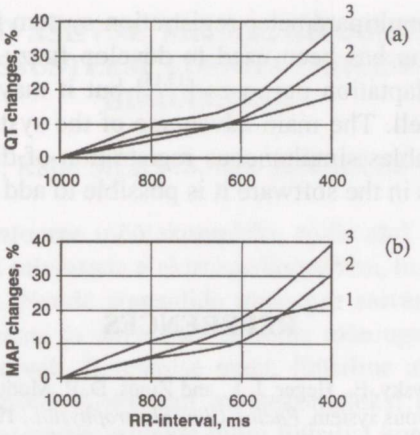


**Fig. 6.** Averaged QT (a) and MAP (b) duration dynamics: curve 1 – spontaneous heart rate (linear correlation coefficient $r = 0.93$ between QT and MAP); curve 2 – heart rate changes during incremental pacing ($r = 0.94$); curve 3 – heart rate changes during fixed rate pacing ($r = 0.98$).

## 4. CONCLUSIONS AND DISCUSSION

Constructed multichannel registration system enables simultaneous registration, digital storing and analysis of ECG and bioimpedance signals in steady state and during the exercise. By analysing these signals it is possible to calculate HR, QT, VV, HI and ACT parameters, which give a physician information about the state of the patient's health. Thanks to simultaneous registration of the signals, it is easy to find out the relations and correlation between the calculated parameters.

Using PC with DAQ and LabVIEW makes the system flexible for future development – it is possible to add additional data channels (12 ECG leads instead of two, for example). All recorded data are stored and can be analysed several times with different programs. LabVIEW graphical environment has a powerful signal processing toolbox; it is easy to develop and integrate new analysing algorithms into the system. At the same time it has elegant and easy-to-use user interface for those who do not have programming experience.

Results of comparison of QT and MAP demonstrate high correlation between these parameters. The correlation does not depend on the character of HR changes – spontaneous or pacing-induced. These findings allow in some cases to use QT, measured from the surface, instead of MAP, which requires endocardial electrogram registration.

We can see that in case of paced HR the character of dynamics of QT as well as MAP duration differ significantly from the parameters at spontaneous HR (Fig. 6). The reason for this fact is not clear yet and needs further investigation.

This noninvasive multiparameter registration system for evaluating patient's physiological condition has been used to develop fuzzy rules for rate adaptive cardiac pacemaker adaptation purposes [13,15] but it may be useful in everyday clinical practice as well. The main advantage of the system is that it is flexible and simple and it enables simultaneous registration of different parameters. By making small changes in the software it is possible to add parameters that need to be measured.

## REFERENCES


1. Browne, K. F., Prystowsky, E., Heger, J. J., and Zipes, D. P. Modulation of the QT interval by the autonomic nervous system. *Pacing Clin. Electrophysiol.*, 1983, **6**, 5/2, 1050–1056.
2. Bexton, R. S., Vallin, H. O., and Camm, A. J. Diurnal variation of the QT-interval – influence of the autonomic nervous system. *Br. Heart J.*, 1986, **55**, 3, 235–258.
3. Lau, C.-P. *Rate Adaptive Cardiac Pacing: Single and Dual Chamber*. Futura, New York, 1993, 73–135.
4. Surawicz, B. Will QT dispersion play a role in clinical decision-making? *J. Cardiovasc. Electrophysiol.*, 1996, **7**, 8, 777–784.
5. Rossi, P., Plicchi, G., Canducci, G., Rognoni, G., and Aina, F. Respiratory rate as a determinant of optimal pacing rate. *Pacing Clin. Electrophysiol.*, 1983, **6**, 2/2, 502–507.
6. Lau, C.-P., Antoniou, A., Ward, D. E., and Camm, A. J. Initial clinical experience with a minute ventilation sensing rate modulated pacemaker: Improvements in exercise capacity and symptomatology. *Pacing Clin. Electrophysiol.*, 1988, **11**, 11/2, 1815–1822.
7. Min, M., Parve, T., Eek, A., and Märtin, H. An instrument for measurement of vector parameters of electrical bio-impedance. In *Proc. IX International Conference on Electrical Bio-Impedance*. Heidelberg, 1995, 40–43.
8. Min, M. and Parve, T. Current mode signal processing in lock-in instruments for electrical bio-impedance measurement. In *Proc. 1st International Conference on Bioelectromagnetism*. Tampere, 1996, 167–168.
9. Ebert, T. J., Eckberg, D. L., Vetrovec, G. M., and Cowley, M. J. Impedance cardiograms reliably estimate beat-by-beat changes of left ventricular stroke volume in humans. *Cardiovasc. Res.*, 1984, **18**, 6, 354–360.
10. Tompkins, W. J. *Biomedical Digital Signal Processing: C-Language Examples for the IBM PC*. Prentice-Hall, New Jersey, 1992, 236–263.
11. Neuvo, Y., Dong, C.-Y., and Mitra, S. K. Interpolated finite impulse response filters. *IEEE Trans. Acoust. Speech Signal Process.*,1984, **32**, 3, 563–570.
12. Alahautala, T. and Lehtinen, R. *Computerized Recognition of the QT Interval from Exercise Electrocardiogram*. TKK Univ. Pr., Tampere, 1993.
13. Lass, J., Hinrikus, H., Kaik, J., and Meigas, K. Measurement of correlation between heart rate and physiological parameters variations. In *Proc. 19th Annual International Conference of the IEEE Engineering in Medicine and Biology Society*. Chicago, 1997, 308–310.
14. Hinrikus, H., Kaik, J., Meigas, K., and Lass, J. Correlation between measured surface and endocardial electrical signals of heart. In *Proc. 2nd International Conference on Bioelectromagnetism*. Melbourne, 1998, 127–128.
15. Lass, J., Meigas, K., and Hinrikus, H. Simultaneous measurement of rate adaptive heart pacemaker sensing parameters. In *Proc. 10th Nordic-Baltic Conference on Biomedical Engineering, Satellite Symposium on Bioelectromagnetic and Biomechanic Measurements*. Tallinn, 1996, 5–6.

# MITTEINVASIIVNE MITMEPARAMEETRILINE SALVESTUSSÜSTEEM FÜSIOLOOGILISE SEISUNDI HINDAMISEKS

Jaanus LASS, Kalju MEIGAS, Hiie HINRIKUS ja Jüri KAIK

On koostatud laboratoorne mõõtekompleks, mille abil on võimalik registreerida ja digitaalsel kujul salvestada elektrokardiogrammi, bioimpedantsi ning keha liigutuste intensiivsust. Nende signaalide üheaegse salvestuse ja analüüsi tulemusel on võimalik hinnata ja omavahel võrrelda mõningaid füsioloogilisi parameetreid, nagu QT-intervall, hingamise maht, füüsiline aktiivsus ning südamelihase impedantsimuutused. Koostatud mõõtekompleksi on rakendatud tervete inimeste uurimiseks, kasutades erinevat tüüpi füüsilisi koormusteste. On kirjutatud salvestus- ja analüüsiprogrammid LabVIEW keskkonnas kasutamiseks.