İSTANBUL TECHNICAL UNIVERSITY ★ INSTITUTE OF SCIENCE AND TECHNOLOGY

MONITORING OF CARDIO RHYTHM WITH ACCELEROMETER (Accelero-Cardio-Gram-ACG) OVER WIRELESS BODY AREA NETWORK

M.Sc. Thesis by
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Programme : Biomedical Engineering

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<u>İSTANBUL TEKNİK ÜNİVERSİTESİ ★ FEN BİLİMLERİ ENSTİTÜSÜ</u>

KALP RİTMİNİN İVME ALGILAYICI İLE TAKİBİ
(İvmesel-Kalp-Ritim-Takibi - İKT)
VE KABLOSUZ VÜCUT ALAN AĞI ÜZERİNDEN İLETİMİ

YÜKSEK LİSANS TEZİ

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FOREWORD

First I would like to thank to my very valuable family and especially to my mother Türkan Güner for her unlimited supports for my education. I would like to express my deep appreciation to my teachers.

This work is supported by ITU Institute of Science and Technology.

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Hasan Basri GÜNER

Electronics & Telecom. Engineer

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ABBREVIATIONS

ACG : Accelero-Cardio-Gram
ADC : Analogue-to-Digital Converter
APS : Application Support Sub-layer
BAN : Body Area Network
COTS : Commercial Off-The-Shelf
CPU : Central Processing Unit
CSD : Circuit-Switched Data
CSM : A/CA Carrier Sense Multiple Access/Collision Avoidance
DAC : Digital-to-Analogue Converter
DSSS : Direct Sequence Spread Spectrum
ECG : Electrocardiogram
EDG : Enhanced Data for GSM Evolution
EDR : Enhanced Data Rate
EEG : Electroencephalogram
EGPRS : Enhanced General Packet Radio Service
EMG : Electromyogram
FFD : Full Function Device
FH : Frequency Hopping
FHSS : Frequency Hopping Spread Spectrum
FSK : Frequency Shift Keying
G1 : The reference for the beginning of the heartbeat in the ACG signal
GGSN : Gateway GPRS Support Node
GPRS : General Packet Radio Service
GPS : Global Positioning System
GPSK : Gaussian Phase-Shift Keying
GSM : Global System for Mobile communications
HCI : Host Controller Interface
HS : Heart Sound
HSCSD : High Speed Circuit-Switched Data
IEEE : Institute of Electrical and Electronic Engineers
IP : Internet Protocol
ISM : Industrial, Scientific, and Medical
İKT : İvmesel-Kalp-Ritim-Takibi
L2CAP : Logical Link Control and Adaptation Protocol
LCP : Link Controller Protocol
LMP : Link Manager Protocol
MAC : Medium Access Control
MEMS : Micro Electro-Mechanical Sensor
MCU : Microcontroller Unit

NMEA : National Marine Electronics Association
NWK : Network layer
OBEX : Object Exchange Protocol
OFDM : Orthogonal Frequency Division Multiplex
OS : Operating System
PAN : Personal Area Network
PC : Personal Computing
PL : Personal Logger
PDA : Personal Digital Assistant
PHY : Physical layer
PSK : Phase-Shift Keying
QoS : Quality of Service
RAM : Random Access Memory
RF : Radio Frequency
RFD : Reduced Function Device
PPP : Point-to-Point Protocol
SAP : Sensor Access Point
SDP : Service Discovery Protocol
SIG : Special Interest Group
SN : Sensor Node
SPI : Serial Peripheral Interface
SSH : Secure Shell
TCP : Transport Control Protocol
TDMA : Time Division Multiple Access
UART : Universal Asynchronous Receiver Transmitter
UDP : User Datagram Protocol
UMTS : Universal Mobile Telecommunication System
UWB : Ultra Wide Band
WBAN : Wireless Body Area Network
WiFi : Wireless Fidelity
WLAN : Wireless Local Area Network
WPAN : Wireless Personal Area Network
ZDO : ZigBee Device Object
ZigBee : A standard wireless technology governs by ZigBee Alliance
ZNC : ZigBee Network Coordinator

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MONITORING OF CARDIO RHYTHM WITH ACCELEROMETER (Accelero-Cardio-Gram-ACG) OVER WIRELESS BODY AREA NETWORK

SUMMARY

Recent technological advances in wireless communications, mobile computation, and sensor technologies have enabled the development of low-cost, miniature, lightweight, intelligent wireless sensor devices. A collection of these devices can be placed strategically on the key positions in the human body and connected by means of a wireless network to form a Wireless Body Area Network.

MEMS accelerometers, gyros, RFs etc. have recently attracted a great deal of attention from researchers both in academy and industry too. This is primarily due to its unique capabilities and promising applications in areas like healthcare, fitness, sports, military and security. In the healthcare domain, MEMS accelerometer promises to revolutionize healthcare system through allowing inexpensive, non-invasive, ambulatory monitoring of human's health-status anytime, anywhere. Our main target is to implant the MEMS accelerometer based ACG sensor to a close location of the heart.

In this thesis, we propose a wireless accelerometer sensor based prototype system for remote monitoring of the cardio rhythm activities. The system consists of IEEE 802.15.4 based wireless RF and MEMS accelerometer couple (sensor node-SN), a sensor access point (SAP) and a PC. The PC collects the ACG signal data collected by the SN over SAP and extracts the G1 feature then calculates the heartbeat rate. After the calculation, the results are served over the PC to the internet for the use of any mobile user.

The ACG sensor measures cardio rhythm with an accelerometer. In the future use, it's intended to make the all calculations in the SN's MCU. And SN will send any information if it's needed. This will make a very clean RF environment. However to make these self calculating SN device, the key is to make the all system ultra low power as possible.

The ultra low power SN could easily be implanted into the patients' body. If the SN is enough low power consuming device then an energy harvester and a micro battery will be very suitable to make a successful implant. In the thesis it will be shown that it is possible to keep the CPU power to calculate the heartbeat rate with the very low overhead algorithms. Therefore we chose the zero-crossing algorithm to analyze the cardio rhythm.

In the past, for the cardio rhythm analysis, zero-crossing algorithm was used very successfully for the ECG signals. However ACG signal is very different than the ECG signal. ACG signal is more like heart sounds. Therefore zero-crossing algorithm had been changed for the efficient results as explained in detail.

Currently the system is in a prototype phase and is developed as a proof-of-concept. The proposed system, once perfected, can be used in many different application scenarios. For example, remote monitoring of the elderly people, people with disabilities, patients undergoing physical rehabilitations, athletes or soldiers during training/exercises, etc. It’s well understood that the method can also be used for the healthy people for the protection purposes because of its implementation with the lowest risk possible. The implant can be made onto the closest bone to the heart (sternum) from the upward of the bone. It means easy surgery and to be far from heart.

A tri-axial accelerometer based signal monitoring and analysis method using wireless sensors technology was designed and developed in this study. In the thesis, this method is named as Accelero-Cardio-Gram (ACG) and the specific feature signal is named as G1 by us. The system provides an application for recording potentially important medical symptoms. The cardio rhythm features are used to detect life-threatening arrhythmias, with an emphasis on the software for analyzing the heartbeat intervals. If any abnormality occurs, over the wireless network any alarm can be sent to the doctor's PC, Personal Digital Assistant (PDA) or Smart Phone over a central server.

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ÖZET

Gelişen kablosuz ağ teknolojileri, mobil işlemciler ve algılayıcı teknolojileri, çok küçük ve ekonomik akıllı kablosuz algılayıcı cihazların yapımına olanak sağlamıştır. Bu kablosuz algılayıcı cihazlar, insan vücudu üzerinde hayati işaretlerin algılanması için bir kablosuz algılama ağı oluşturmak üzere bir araya getirilmiştir ve böylece Kablosuz Vücut Alan Ağı (KVAA) ortaya çıkmıştır.

MEMS teknolojisi ile ivme algılama, açı algılama, RF, v.b. hem akademik hem de endüstriyel araştırmalarda çok önemle üzerinde durulan bir konu haline gelmiştir. Bunun başlıca sebepleri, MEMS teknolojisi ile oluşturulabilecek çok sayıda sağlık, kontrol, kişisel bakım, spor, askeri amaçlı kullanım ve güvenlik gibi alanlarda sayısız uygulama olanağı bulmuş olmasındandır. Sağlık alanında ise devrimsel nitelikte bir gelişim gösteren MEMS ivme algılama sistemleri, ucuz, vücut ile direk temas etmeyen yapısı ile acil durum izleme ve hayati işaret takibinin her zaman ve her yerde yapılmasına olanak sağlayacak yeni uygulamalara olanak vermiştir. Tezin amacı kalbe yakın bir yere implante edilebilecek MEMS ivme algılayıcı tabanlı kablosuz bir İKT algılayıcısı sistemi yapmaktır.

Bu tezin konusu içerisinde, kablosuz ivme algılayıcı tabanlı bir prototip ile uzaktan kalp ritmi takibi yapılacaktır. Sistem IEEE 802.15.4 tabanlı kablosuz bir RF ve ivme algılayıcısı ikilisinden oluşan bir Algılayıcı Nokta (AN), bu algılayıcı noktadan İKT sinyallerini toplayan bir algılayıcı erişim noktası (AEN) ve bir PC'den oluşmaktadır. PC İKT sinyallerini AEN üzerinden AN ile toplar ve İKT sinyalinden G1 kompleksini ayırdeder ve bunun sonucunda kalp ritim belirlemesi hesaplaması yapar. Hesaplama sonuçları internete bağlı bir sunucu üzerinden ilgililerin PC'lerine, PDA'lerine veya tarayıcı çalıştırabilen mobil telefonlarına iletir.

İKT algılayıcısı temel olarak ivme algılayıcısı ile kalp ritmini ölçmektedir. İlerisi için hedeflenen çalışma ise, PC üzerinde geliştirilmiş ve verimli hale getirilmiş programın AN içerisine gömülmesidir. Bu şekilde tüm hesaplamalar AN üzerinde yapılıp bir acil durumda veya talep olduğunda AN'den bilgi gönderilecektir. Bu sayede sürekli RF haberleşmeden kaynaklanan bir kirlilik olmayacaktır. Ancak bu hedefe ulaşabilmenin anahtarı ise AN üzerinde olabilecek en düşük güç tüketimini elde edebilmektir.

Düşük güçlü bir AN kolayca insan vücudu içerisine yerleştirilebilir ve bir mikro pil ve bu pili şarj eden vücut içi enerji toplayıcıları ile teorik olarak sonsuza kadar çalıştırılabilirler. Tezde düşük güç harcamasını sağlayacak kısa bir kalp ritim analiz algoritmasının nasıl geliştirileceği gösterilmiştir. Mikroişlemci üzerine çok az yük getiren bir kalp ritim analiz yöntemi olarak sıfır-geçişleri yöntemi ile algoritma gerçekleştirilmiştir.

Sıfır-geçişleri yöntemi ile çok başarılı olarak EKG analizi yapılabilmiştir. Ancak İKT sinyalleri EKG sinyallerinden farklıdır ve daha çok kalp ses sinyallerine benzemektedir. Bu sebeple daha verimli bir program geliştirmek için klasik manada bilinen sıfır-geçişleri algoritması İKT sinyallerine uygun olarak değiştirilmiştir.

Şu anda tez konusu sitem prototip aşamasındadır ve sadece kavramların kanıtlanması anlamında gerçeklenmiştir. Olgunlaştırıldıktan sonra bir çok farklı uygulamada kullanılabilecektir. Örneğin, yaşlılar, özürlüler, fizik tedavi görenler, atletler, askerler v.b. Görüldüğü gibi bu metod sağlıklı insanlarda, koruyucu sağlık anlamında da kullanılabilir. Bunun sebebi ise kalbe yakın bir kemiğe (sternum), yukarıdan ve kolay bir cerrahi ile implante edilebilmesindendir. Yani kalbe olabildiğince uzakta kalınarak düşük riskli bir yerleştirme operasyonu yapılabilecektir.

Üç-boyutlu ivme algılayıcısı temeline dayalı takip ve analiz yöntemi kablosuz haberleşme yöntemi ile birlikte tasarlanmış ve geliştirilmiştir. Tezde bu yönteme; İvmesel-Kalp-Ritim-Takibi veya Accelero-Cardio-Gram (ACG) ismi, ritim belirleyici sinyal bölümüne ise G1 ismi verilmiştir. Sistem tibbi açıdan çok önemli verilerin toplanmasını sağlamaktadır. Kalp ritmi analizi, kalp atış aralıklarının hesaplanması ile durum analizi yapılması şeklinde özetlenebilir. Tesbit edilen herhangi bir bozukluk kablosuz ağlar üzerinden mobil cihazları olan doktor veya sağlık personeline gönderilir.

1. INTRODUCTION

The continuous monitoring of the vital signs has the potential to improve the quality of patients' lifes who have chronic diseases. Especially the growing metabolic syndrome wave is addressing the monitoring of the patiens everywhere. Recent advances in sensor technology allow continuous, real-time ambulatory monitoring of multiple physiological signals including; Heart Sound (HS), Electrocardiogram (ECG), body temperature, respiration, blood pressure, oxygen levels, and glucose levels. These systems are conveniently packaged as a single product and could be used to give a complete picture of the patient health. They generate large amounts of patient data that must be intelligently analyzed and archived to be most useful. However they are not small size and implantable.

The continuous monitoring would allow telecare systems to accommodate a larger number of pathologies including asymptomatic conditions like atrial fibrillation for which intermittent monitoring is not sufficient. Technology that would allow healthcare providers to deploy, configure, and manage such monitoring systems, would provide a tremendous service to the healthcare industry while at the same time improving the quality of life for thousands of patients.

In the thesis, we develop a robust platform for real-time monitoring of patients when they are staying in their home or when they are mobile. Then the system transfers the medical data to the doctors who are working at the hospitals or at the call centers. The system is accelerometer based cardio rhythm analysis system and in the thesis we called the method as ACG, Accelero-Cardio-Gram. The signal that we analyze was named as ACG and the G1-complex is the signal section of our interest.

In the thesis, a cardio rhythm analysis system will be realized by a MEMS accelerometer sensor which will be used to collect ACG data over the pressure sensitive chest bone (sternum) and the wireless communication method will be IEEE 802.15.4. The zero-crossing algorithm will be modified to use for the ACG signals. The reason why this algorithm was chosen is its low calculation overhead resulting low power consumption and low CPU time occupation. Thus one of the main objectives of the thesis is to make a low power algorithm to analyze the heart beat rate together with the low power sensor and a low power communication technology to be able to implant the system into the body in connection with the energy harvesters.

1.1 Cardio Rhythm Analysis Algorithms in Literature

For the purpose of heartbeat rate analysis with ECG, the Pan-Tompkins analysis method is widely used. Another popular way is using wavelet analysis. Beside these methods there are several named methods however they all have the same basic maximum-minimum finding approach.[4]. In the thesis, a much practical, less memory usage and less cpu power consuming “zero-crossing method” will be used.

Wavelet is time-limited signal that interferes with components of the signal to be analyzed [2]. The wavelet and its window function (that makes the wavelet time-limited) is carefully designed to interfere with components of interest in the measured signal [15]. Convolution is performed between the wavelet and the signal. This is a very suitable method for the medical signal analysis with the accelerometer. However it is out of the scope of this thesis because of its highly power consuming calculation load.

The Pan-Tompkins algorithm proposes a real-time QRS detection based on analysis of slope, amplitude, and width of QRS complexes. It includes a series of filters and methods that perform low pass, high pass, derivative, squaring and integration procedures. Filtering reduces false detection caused by the various types of interference present in the ECG signal. This filtering permits the use of low thresholds, thereby increase the detection sensitivity. The algorithm adjusts the thresholds automatically and parameters periodically to adapt to changes in QRS morphology and heart rate.

The heart rate variability analysis has great importance in medical diagnosis. In most cases, the temporal location of the R-wave is taken as the location of the QRS complex and S1-wave is taken as the reference of the heart beat location of the HS complex wave forms like the G1 complex of the ACG signals. Missed or falsely detected beats are problematic in all of these applications and may lead to poor results. Detection errors can be reduced by the application of computationally more expensive algorithms, for instance by the implementation of reverse search methods. However, particularly in the case of battery-driven devices, the computational complexity needs to be kept low. Hence, a trade off between computational complexity and detection performance needs to be found.

The detection of QRS complexes and R-waves in ECG signals has been studied for several decades. Most of the earliest algorithms are based on feature signals obtained from the derivatives of the ECG signal. As long as no additional rules for the reduction of false detections are applied, these methods are characterized by low computational complexity and relatively poor detection results in the presence of problematic signals (e.g., containing baseline drift, noise and artifacts, as well as changes in the QRS morphology). The proposed method is based on a feature that is obtained by counting the number of zero crossings per segment. It is a feature signal that is largely independent of sudden changes in the amplitude level of the signal and is robust against noise and pathological signal morphologies. It is shown that this feature can be used for a computationally simple algorithm with a high detection performance [16].

This thesis is concerned with the design and implementation of a Wireless-accelerometer sensor based prototype system for monitoring mobile user's Cardio rhythm by an accelerometer sensor.

1.2 Background of the Used Technologies

In the recent years, there have been tremendous advances in wireless communications, very small and precise MEMS (Micro Electro-Mechanical Senor) accelerometer technology, ultra low power mobile computing, energy harvesting power supplies and electronics manufacturing industry. These technological progresses have led to the emergence of a new generation of intelligent wireless sensor devices which are very small in size, light in weight, yet smart and powerful in functionality. Each sensor device is typically capable of sensing, sampling, processing and communicating one or more physical, physiological, or biological signals from the environment where they are deployed. A number of these wireless sensor devices can be seamlessly integrated into a Wireless Body Area Network (WBAN) to construct a wireless system for monitoring human body parameters such as body motions, body temperature, heartbeat rate, brain activities, respiration, blood oxygen saturation, blood sugar level, etc. These sensor network on the body may be realized as wearable or implementable. The driving forces include increasing number of elderly people and chronically-ill patients who need long-term care and continuous monitoring. According to UN statistics, the worldwide population of those over 60 is predicted to reach two billion by 2050 [21]. Assuming current trends continue, this century will see the first time in human history that the old outnumber the young.

It will be important in the coming years to develop technology which can reduce the work-load on the care-givers. On the other hand, the traditional monitoring systems fall short in many ways. These systems typically use equipments which are very costly, bulky, and unconformable to use due to wiring and cables which restrict user's movement and obstruct their normal activities. Besides, they often require the user to stick into the place where the monitoring equipments are deployed. For example, patients need to stay in hospital for the duration of the monitoring sessions. Thus, too much resource, efforts and costs can get wasted (e.g., dedicated beds, dedicated healthcare staffs, costs to stay in hospital etc). In addition, monitoring is typically discontinuous and limited to a specific period of time, which only gives a "snapshot" of the health status. Thus, transient anomalies may go undetected.

IEEE 802.15.4 based wireless body monitoring systems and MEMS accelerometer sensors have a number of advantages over the traditional wire-based monitoring systems: Wireless systems utilize low-cost, tiny, lightweight, wireless sensor devices which do not interfere with (or restrict) user's normal activities. With wireless systems, a user can be monitored at anytime, anywhere, and for any duration. It can be used to implant sensors in the human body. MEMS sensors have very low sizes suitable to implant and they have considerably low power consumptions to operate with the in-body energy harvesters [8].

1.3 Motivation of the Thesis

The target of the thesis is to design a cardio rhythm analysis system with an accelerometer sensor and sending the information to a remote point over a wireless communication module. The system is developed as the preparation for the safe and reliable implant onto the chest bone named sternum. The prototype system could be used to monitor mobile user's or patient's cardio rhythm via IEEE 802.15.4 standard based wireless module.

If a very small size and an ultra low power MEMS accelerometer sensor are used to capture a very clear heartbeat pressure wave signal and if this sensor system can be combined with an ultra low power and also very small size wireless communication IC, this final system can work with a micro-battery and an in-body energy harvester charger, thus, the system which was called sensor node (SN) can make a cardio rhythm calculation in its embedded processor and can transmit the results on demand or in case of emergency. Thus we may keep the transmitting data payload as low as possible from a wireless implant system to outside located sensor access point (SAP). For this target, we will show the modified version of the zero-crossing algorithm which can make a cardio rhythm monitoring with very short calculations to save considerable power with an accelerometer sensor on the PC. It will be seen that the system can be used to implant into the body to monitor cardio rhythm.

After the brief introduction to the thesis as finalized here, the study will be classified as the following chapters:

In Chapter 2; the nature of the ACG signal and the G1 complex section of the signal will be defined to extract the feature signal to use for the temporal localization of the feature signal. Then the technologies used will be explained.

In Chapter 3; after the short discussion on the current algorithms used to monitor heart rate, the architecture of the realized test system will be explained.

In Chapter 4; the processes of the cardio rhythm analysis will be explained in detail to explain the principles of the cardio rhythm analysis software.

In Chapter 5; the sensitivity of the proposed system will be calculated and discussed with the numerical results and the future expectations will be summarized.

2. THE ACCELERO-CARDIO-GRAM (ACG) SIGNALS AND THE TECHNOLOGIES USED TO ANALYZE THE CARDIO RHYTHM

2.1 ACG Signals and the G1 Complex

The sampling frequency of of the ACG sensor node (SN) was set to 100Hz for the examples of the thesis. The most smilar signal to the ACG sensed signal is the heart sound (HS) signal. The heart sounds (HS) are the heart originated pressure waves. And the pressure is proportional to the force so acceleration. The following figure shows how heart sounds come out [17].

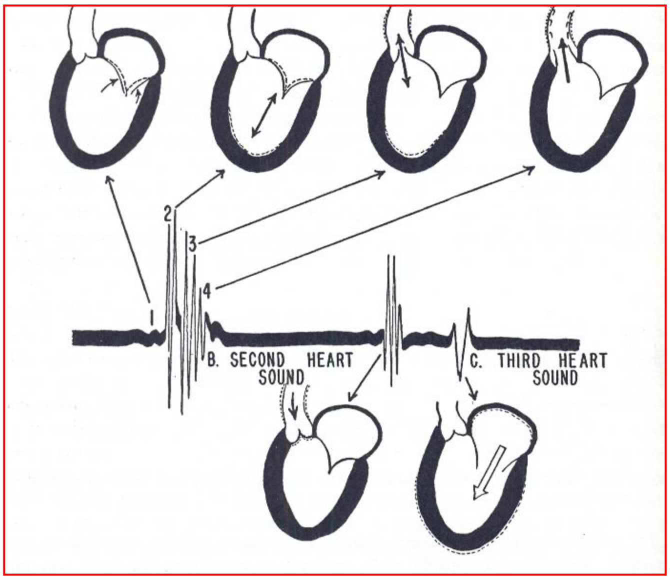

Figure 2.1: Genesis of the heart sounds.

Consequently, we may use HS to smilirize the acceleration shape and location. However the difference is coming from the location that the acceleration sensor was implemented and the response of the accelerometer system. Furthermore, there is not any effective usage of HS for the cardio rhythm analysis.

The spectrum of the heart sound signals contains a band of frequencies in the range of 5–500 Hz. The spectrum of S1 and S2 components can extend up to the frequencies in the range of 20–200 Hz. The start of a heart cycle, systole, corresponds to the QRS complex on the ECG [10]. S1 occurs at the end of the isometric contraction period during systole and S2 occurs after the iso-volumetric relaxation period during diastole. The timing of the heart sounds can be seen in Figure 2.2.

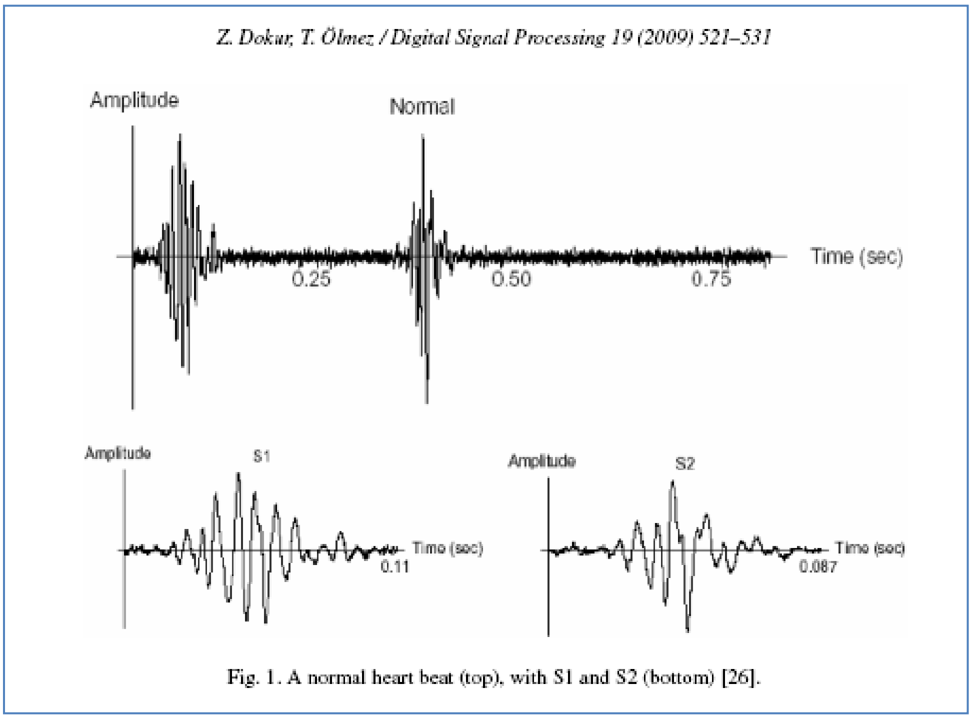

Figure 2.2: S1 and S2 complexes timings.

An electrocardiogram (ECG) represents the electrical current moving through the heart during a heartbeat. Each heartbeat begins with an impulse from the heart's pacemaker (sinus or sinoatrial node) [17]. Hence the electrical signal waveform is much different than the pressure and vibration based signal of the ACG sensor.

The signal drawn in Figure 2.3 is the ACG signal. And the highlighted part is the G1complex of the ACG signal. After the definition and the comparison of the ACG signal and the G1 complex these terms and signal featuree will be used to analyze the cardio rhythm in the thesis.

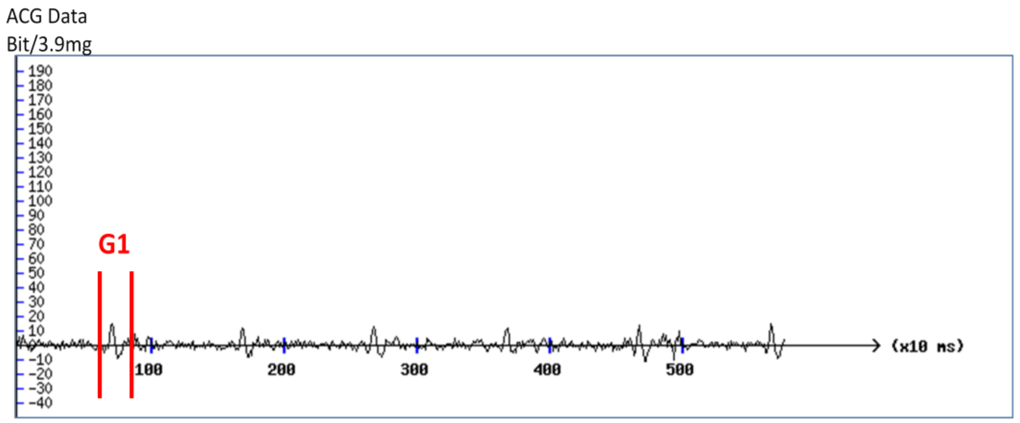

Figure 2.3: G1 complex at the same location with S1.

Above mentined G1 complex of the ACG is at the same place with the S1 of the heart sound signal (HS). G1 and S1 are formed by the same resources. In the thesis, the designed system is extracting the G1 complex to analyze heart beat rate. Because of the smilar pressure wave nature of the both ACG and HS signals, S1 and G1 is going to be used in place of each other to describe the G1 signal.

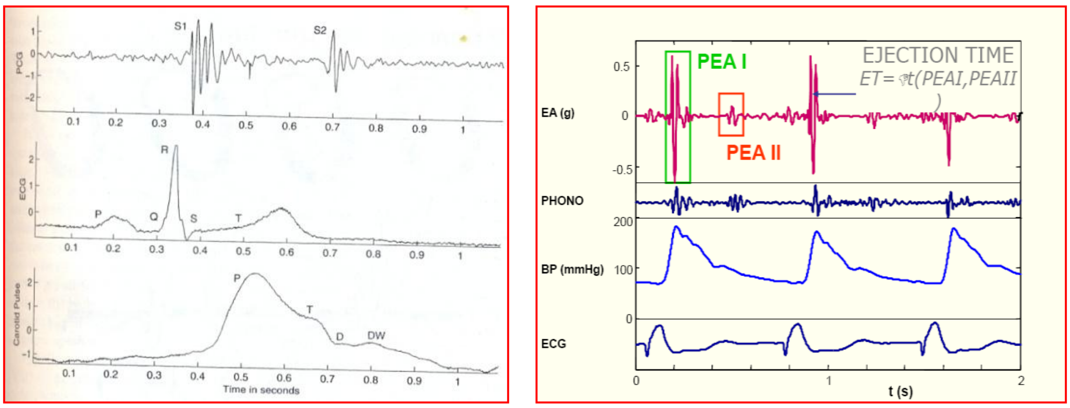

Figure 2.4: Simultaneous PCG, ECG, Carotid Pulse, Endocardial Acceleration and HS.

The high-pass cut-off frequency of the G1 complex is 20 Hz for the ACG. The high frequency component is insignificant in this experiment then the lowpass part of the bandpass filter was omitted.

The only noise considered in the ACG signal is the body activity. The frequency of the noise in the signal is less than 10 Hz in the thesis. Therefore the gravity component is dynamically changing with the maximum 10 Hz frequency i.e. the fixed 1g amplitude on the each axis of the sensor. The highpass filter cut-off frequency (20 Hz) is quite well positioned to eleminate the DC component for the static measurement and the diynamic effects of the gravity force for each axis [3].

2.2. Placement of the ACG Sensors to Collect Data

We use the accelerometer to capture the heartbeat pressure waves. Especially it affects the closest chest bone structure (Sternum) [12]. The place of the accelerometer should not absorb the vibration sourced by the heartbeat in any position of the subject and it should be exposed to the heartbeat pulses directly.

We collected the acceleration data non-invasively from the several healthy people to use in the thesis. The accuracy would be much higher with the better posining of the accelerometer when it was implanted because there would not be any absorber like skin or any soft tissue under it.

When it was installed on the bone as implant, because of the much higher resonance frequency of the sternum bones than the vibrations of the heartbeat pressure waves, the ACG sensor would sense the ACG signal very clearly.

2.3 Possible Noise Resources in the Measured ACG Signal

There are several filtering techniques to eleminate the out-band signals. We may also make the filtering of the in-band signals with the fixed reference points as an optional method. Generally, the normal needs of a potential heart disease patient can be met with the bandpass filtering. And most of the effects are momentary. However, a better filter should eleminate the following effects according to the use case[17].

- Personal Activity Noise: Body Activity, Breathing, Sneezeing, Coughing, Running, Falling Down, Interior Organs' Sound Effects.
- Environmental Noise: Vehicles: Cars, Trains and Planes, their motions and vibrations.
- Intervening Devices: Ultrasound, MR and Other Possible Medical Devices which can form a mechanical effection or vibration.

2.4 The Accelerometer Technology Used for the ACG Signal

The MEMS technology is a very fastly growing technology and for the accelerometer used in the thesis we prefered the MEMS technology because of its superior advantages to make an implant e.g. its growing accuracy, its ultar small size, its energy efficiency and its low cost. There are several MEMS accelerometer manufacturing technologies. The “Surface Micromachining” is the latest technology and invested very widely. Therefore we choose the accelerometer manufactured by surface micromachining in the thesis [9].

Surface Micromachining builds devices up from the wafer layer-by-layer. A typical Surface Micromachining process is a repetitive sequence of depositing thin films on a wafer, photopatterning the films, and then etching the patterns into the films. In order to create moving, functioning machines, these layers are alternating thin films of a structural material (typically silicon) and a sacrificial material (typically silicon dioxide). The structural material will form the mechanical elements, and the sacrificial material creates the gaps and spaces between the mechanical elements. At the end of the process, the sacrificial material is removed, and the structural elements are left free to move and function.

For the case of the structural level being silicon, and the sacrificial material being silicon dioxide, the final "release" process is performed by placing the wafer in Hydrofluoric Acid. The Hydrofluoric Acid quickly etches away the silicon dioxide, while leaving the silicon undisturbed. The wafers are typically then sawn into individual chips and the chips packaged in an appropriate manner for the given application. Surface Micromachining is suitable for applications requiring more sophisticated mechanical elements.

In the thesis, the technology of the acceleorometer is the polysilicon surface-micromachined structure built on top of a silicon wafer. Polysilicon springs suspend the structure over the surface of the wafer and provide a resistance against forces due to applied acceleration. The size is less than 0.58 mm x 0.66 mm. Deflection of the structure is measured using differential capacitors that consist of independent fixed plates and plates attached to the moving mass. Acceleration deflects the proof mass and unbalances the differential capacitor, resulting in a sensor output whose amplitude is proportional to acceleration. Phase-sensitive demodulation is used to determine the magnitude and the polarity of the acceleration [20].

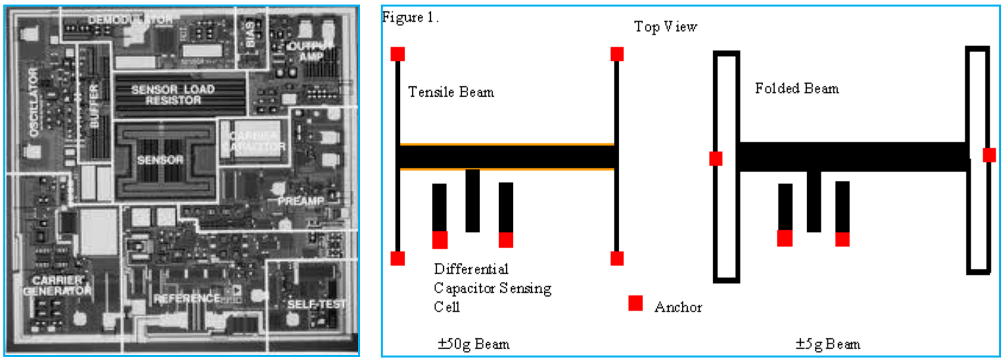

Figure 2.5: MEMS accelerometer IC and the measurement principle.

Since the MEMS technology does not have a long history most of the apllications are under development and we can list some of the known and experienced applications e.g. Body Activity Detection for the Training, Body Position Detection for the Medical Devices, Cardiopulmonary Resuscitation – CPR, Respiration Devices, Phonocardiogram (PCG), Carotid Pulse (CP), Vibromyogram (VMG), Vibroarthogram (VAG). The cardio rhythm analysis with the MEMS accelerometer and this application is not studied before according to our researches and there is not an implant study with it.

2.5 The Wireless Technology Used for the Data Access

In the thesis, the wireless communication is a network consists of several small ACG sensors implantable into the human body like a very basic wireless body area network [3]. The sensors communicate wirelessly through using a known wireless communication technology named ZigBee. ZigBee is a low power, low cost, low data rate, short range wireless technology which built on top of the IEEE 802.15.4. The primary goals of ZigBee are simplicity, long battery life, advanced networking capabilities, reliability and low cost. Some of the applications of ZigBee include; home automation, industrial, remote control, smart tags, sensor networks, medical, and monitoring applications [23].

The IEEE 802.15.4 PHY uses Direct Sequence Spread Spectrum (DSSS) to fight against the potentially high interference levels in the unlicensed frequency bands used. Two physical layers are defined depending on the frequency band: the 868/915MHz PHY and the 2,450MHz PHY. The Table 2.1 shows some of the characteristics including maximum data rate and geographical coverage [11].

As shown in the Table 2.1, the 2.4GHz PHY allows higher data rate than the other two and also it is available worldwide. This makes it more suitable for WBAN applications than the other two options. However, using 2.4GHz ISM frequency band has its own drawbacks, the considerable body attenuation and the potentially high interference level, mainly due to the co-existence of other wireless technologies on this band, most notably IEEE 802.11b (WiFi) and 802.15.1 (Bluetooth).

As shown in the Figure 2.1, ZigBee builds a Network (NWK) layer and an Application (APL) layer on top of the IEEE 802.15.4 MAC and PHY layers. The PHY layer provides the basic communication capabilities of the physical radio. The MAC layer provides services to enable reliable single-hop communication links between devices. The NWK layer provides routing and multi-hop functions needed for creating different network topologies. The application layer includes an Application Support Sub-layer (APS), the ZigBee Device Object (ZDO), and the ZigBee applications defined by the user or designer. Whereas the ZDO is responsible for overall device management, the APS provides servicing to both ZDO and ZigBee applications [11].

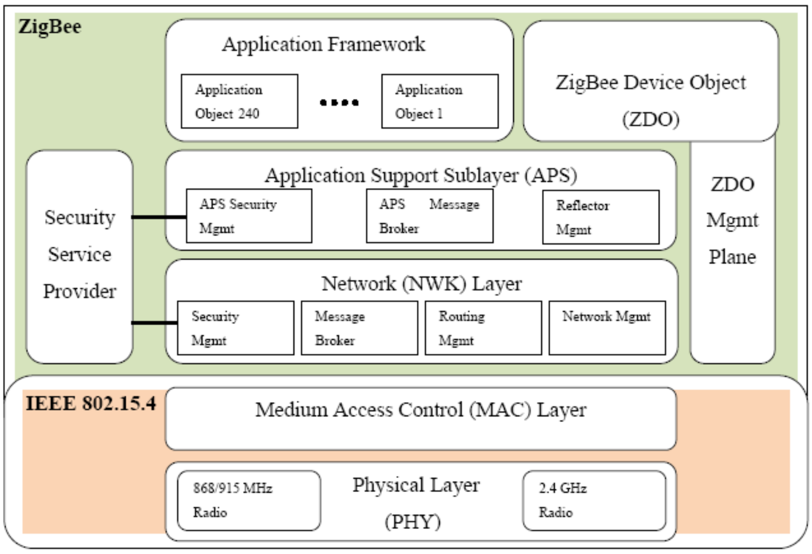

Figure 2.6: ZigBee stack organization.

Reduced Function Device (RFD) and Full Function Device (FFD). RFDs implement a subset of the IEEE 802.15.4 and cannot act as coordinator. FFDs have a full implementation of IEEE 802.15.4 and they can be configured as end-devices, coordinators, or routers in a WBAN [7].

In summary, ZigBee is a low-cost, low-power, low-rate, short-range wireless personal area network (WPAN) technology. ZigBee radio operates in ISM frequency band within three different frequency ranges, 686MHz, 915MHz, and 2.4GHz, and supports data rates of 20kbps, 40kbps, and 250kbps respectively. ZigBee radio employs DSSS to combat interference and fading. It supports security at Link and Network layers. Three networking topologies are supported: star, cluster tree and mesh. It allows up to 64K (65536) nodes in a single network. ZigBee is based on IEEE 802.11.4 MAC and PHY standards which are highly optimized for power consumption. The major drawback of ZigBee, however, is its relatively low data rate.Table 2.1 shows a general comparison of the most modern wireless technologies.

Table 2.1: Comparison table of the possible WBAN technologies.

	WiFi IEEE 802.11b	Bluetooth IEEE 802.15.1	IEEE 802.15.4			
						ZigBee
Application Focus	Web, E-mail, Video	Cable Replacement	Monitoring and Control			
Frequency band	2.4GHz	2.4GHz	2.4GHz (Global)	915MHz (US)	868MHz (EU)	
Spectrum Spreading (SS)	DSSS	Frequency Hopping (FHSS)	Direct Sequence Spread Spectrum (DSSS)			
#channels/schemes	11(US), 13(EU)	10	16	10	1	
Max Data Rate	11Mbit/s	1Mbit/s	250kbit/s	40kbit/s	20kbit/s	
Range (meters)	1 - 100	1 – 10+	1 – 100+			
Network Topology	Star, peer-to-peer	Star	Star, peer-to-peer			+ mesh
Network Size	32	8	64K			
Network join time	<3s	<10s	<<1s			
Real-time support	No	No	Guaranteed time slots			No
Protocol complexity	Medium	High	Simple			Low
System resources	1MB+	250KB+	4KB – 32 KB			
Security	Authentication, encryption	Authentication, encryption	Authentication, encryption			
Power consumption	400-700mW	200mW	60-70mW			
Battery life (days)	0.5 - 5	1 - 7	100 – 1000+			
Success metrics	Speed, Flexibility	Cost, Convenience	Reliability, Power, Cost			

WBAN is an emerging enabling technology with a broad range of potential applications and use cases in diverse application domains including medical, fitness and wellness management, military, safety and security, sports, social networking and entertainments. In medical domain, a WBAN of medical sensors can be used in different scenarios, for example, for sleep staging, for computer-assisted physical rehabilitation, for monitoring patient at home, at hospital, or anywhere [7]. WBAN technology is still in its infancy and there is a lot of research going on. As the technology continues to evolve, new areas of applications will likely emerge withing next few years [18]. Due to the data rates that ZigBee is operating, EDGE is the best matching GSM standard for the remote side communication [19].

3. T HE ARCHITECTURE OF THE CARDIO RHYTHM ANALYSIS SYSTEM

The main cardio signal analysis concept of the thesis base on the “zero-crossing” method. There are different very well defined cardio signal analysis methods widely in use. For example, the classical Pan-Tompkins analysis method is commonly used for the ECG signal analysis. The ECG system itself is a high power consumer system beside this hardware the Pan-Tompkins algorithm or any other derivation based algoritms are consuming too much power too. Therefore Pan-Tompkins like algorithms are not suitable for the battery powered devices since they need long derivation calculations. The wavelet analysis methods have the same calculation overhead difficulties. Thus, the zero-crossing algoithm on the ultra low power 8051 MCU had been used in the system. The name of this device called “sensor node” (SN). This device would be the data collector as well as the data processor. However, for the sake of clarity, the cardio rhythm analysis software had been developed on the PC. Once the program was developed in the most efficient way, to embed it into the SN would be very practical. Therefore we used the SN for the data collection purpose and then we processed the data and served the results on the PC. To collect the data from the sensor node, a sensor access point was used. Thus the system harware consists of three parts, “Sensor Node” (SN), “Sensor Access Point” (SAP) and a “Personal Computer” (PC) [19].

3.1 The Developed System Architecture

The system consists of a data collector node SN to collect the ACG data, quite capable acces point SAP and a PC. The SAP is capable of managing a number of wireless sensor nodes if they work as the same way like the others implanted into the body. The SAP is connected to the PC by a USB connection. The SAP manages its communication by a ZigBee application based on the IEEE 802.15.4 standard. In this thesis the system will be realized by one node for the sake of the prove-of-concept. The ZigBee technology is used to implement the WBAN for the future use and the network has a star topology where all the sensor nodes communicate with an SAP which is configured as ZigBee coordinator. The sensor node incorporates with an MEMS accelerometer. The sensor sends measurement data to the SAP. The SAP collects the data and transmits to the PC to processe them with the cardio rhythm analysis software then finally serves them on the internet. The software was developed on the PHP because of its platform independent and browser based structure. Therefore any PC or mobile device connected to the internet can monitor the patient or subject by using the browsers.

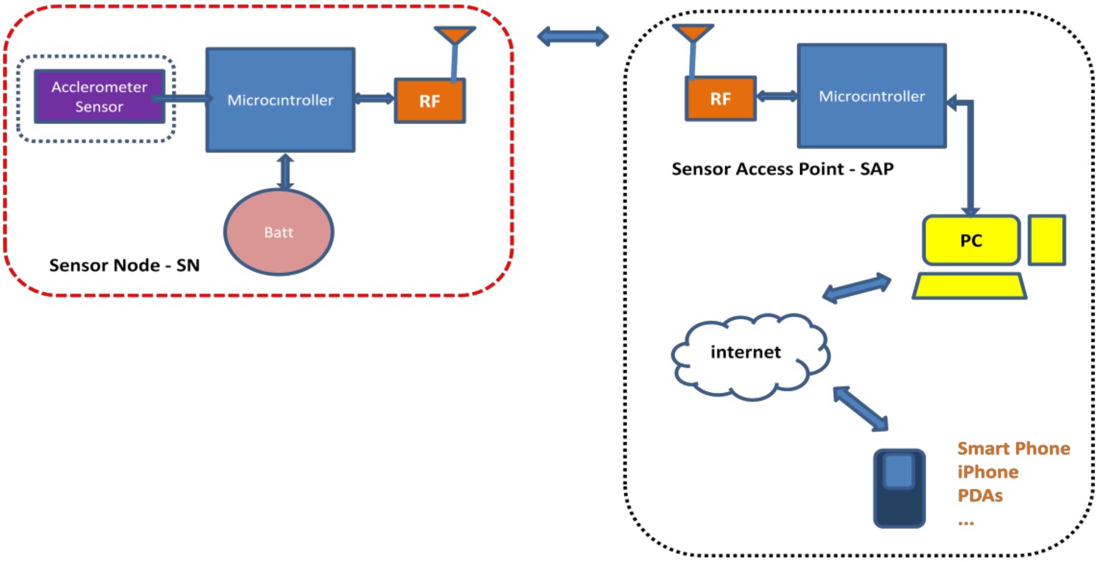

Figure 3.1: The architecture of the developed system.

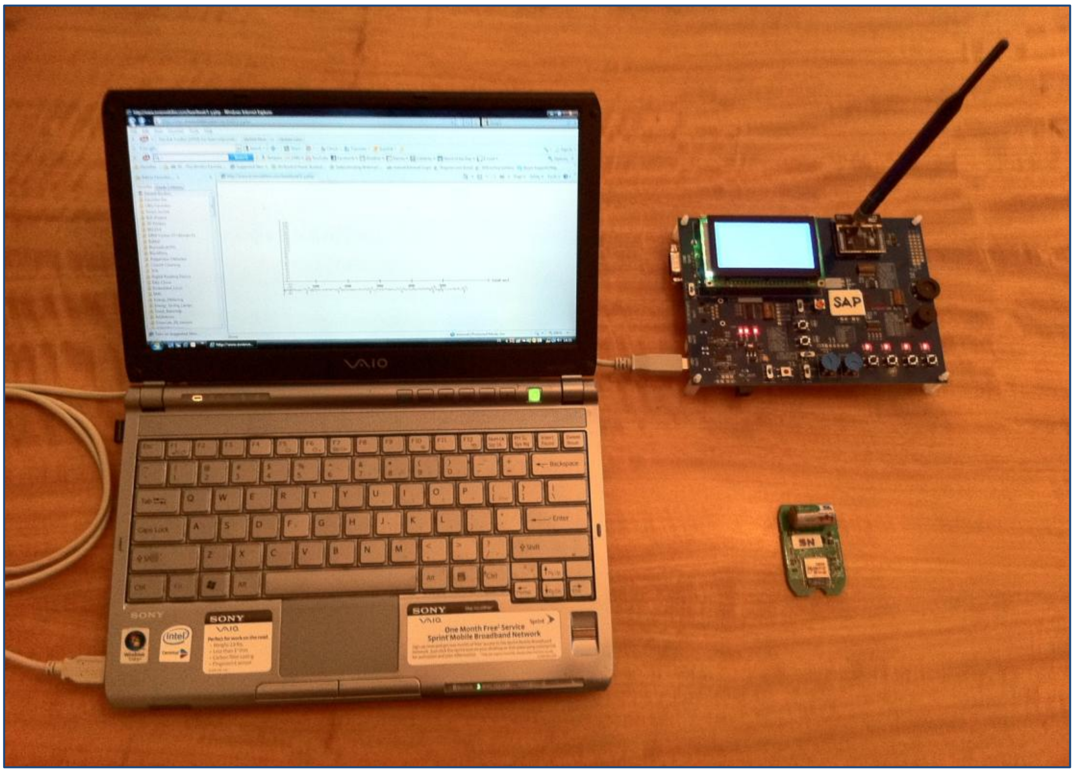

Figure 3.2: Testbed.

3.2 The Sensor Node (SN)

The sensor node will collect ACG data to extract the G1 complexes and then these data will be transmitted to a sensor access point (SAP). The sensor node (SN) was based on the Telecon RP100 ZigBee module which has a 80C51 based MCU. The PCB is desiged using a simple 2-layered structure. The sensor node incorporates with the Analog Devices ADXL345 MEMS accelerometer sensor which is called ACG sensor in the thesis. The ACG sensor can be used for monitoring body motions and gesture recognition as well. Figure 3.2 shows the ACG sensor schematic.

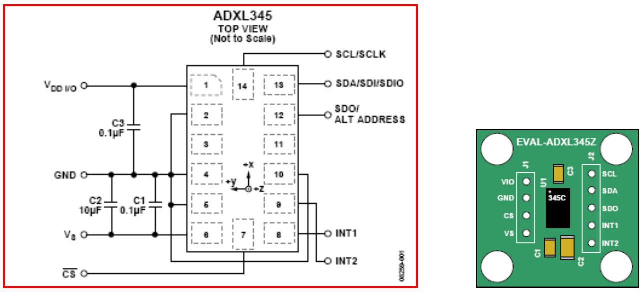

Figure 3.3: ADXL345 accelerometer sensor as the ACG sensor.

The ADXL345 accelerometer used in the thesis has the following sensitivity and noise characteristics as shown in Figure 3.3 and Figure 3.4.

Table 3.1: ADXL345 ACG sensor sensitivity characteristics.

SENSITIVITY	Each axis				
Sensitivity at X_{OUT}, Y_{OUT}, Z_{OUT}	All *g*-ranges, full resolution	230	256	282	LSB/*g*
	±2 *g*, 10-bit resolution	230	256	282	LSB/*g*
	±4 *g*, 10-bit resolution	115	128	141	LSB/*g*
	±8 *g*, 10-bit resolution	57	64	71	LSB/*g*
	±16 *g*, 10-bit resolution	29	32	35	LSB/*g*
Sensitivity Deviation from Ideal	All *g*-ranges		±1.0		%
Scale Factor at X_{OUT}, Y_{OUT}, Z_{OUT}	All *g*-ranges, full resolution	3.5	3.9	4.3	mg/LSB
	±2 *g*, 10-bit resolution	3.5	3.9	4.3	mg/LSB
	±4 *g*, 10-bit resolution	7.1	7.8	8.7	mg/LSB
	±8 *g*, 10-bit resolution	14.1	15.6	17.5	mg/LSB
	±16 *g*, 10-bit resolution	28.6	31.2	34.5	mg/LSB
Sensitivity Change Due to Temperature			±0.01		%/°C

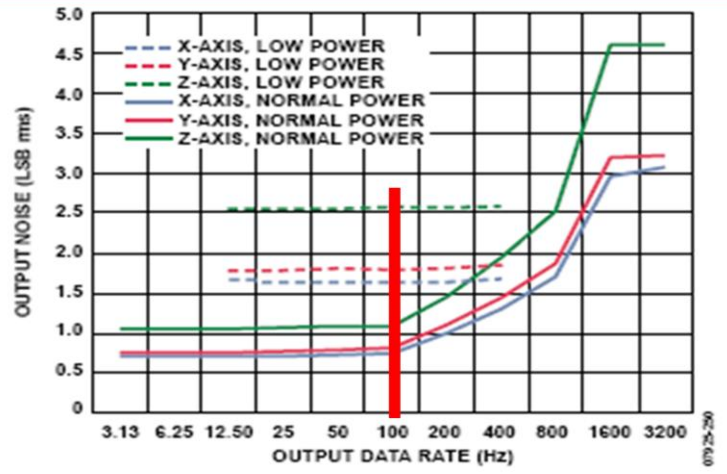

Figure 50. Noise vs. Output Data Rate for Normal and Low Power Modes, Full-Resolution (256 LSB/g)

Figure 3.4: ADXL345 ACG sensor noise characteristics.

Each sensor node contains the following hardware components:

- Sensing unit: ADXL345 three-axis accelerometer
- Power supply unit: 3V coin-cell Lithium battery
- Microcontroller Unit (MCU): 80C51 Generic version.
- Communication unit: RP100 ZigBee module.
- Memory/Storage unit: 2Kbyte on-chip RAM memory, and 64Kbyte on-chip FLASH.
- Radio: 2.4 GHz IEEE 802.15.4 standard.

The following schematic and the PCBA picture are of the sensor node (SN).

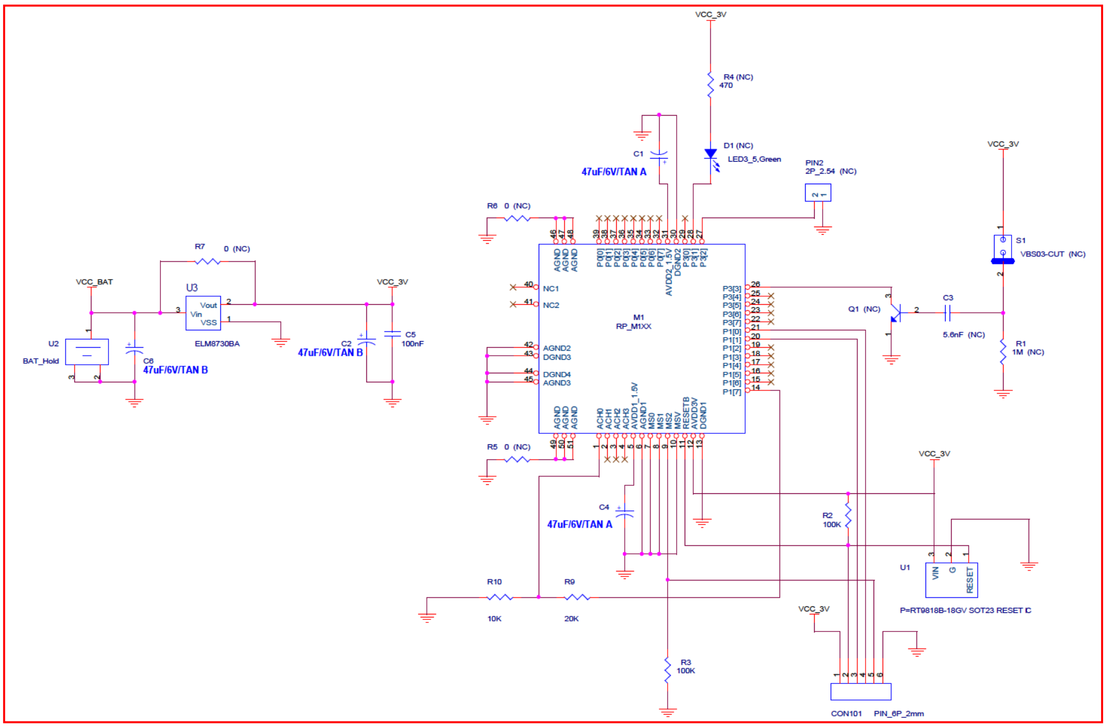

Figure 3.5: Sensor Node (SN) schematic.

Figure 3.6: Sensor Node (SN)

The ADXL345 is a three-axes digital output linear accelerometer that includes a sensing element and an IC interface able to take the information from the sensing element and to provide the measured acceleration signals to the external world through an I2C/SPI serial interface. The sensing element, capable of detecting the acceleration of the heart pressure waves, is manufactured using a dedicated process developed by ADI to produce inertial sensors and actuators in silicon. The IC interface instead is manufactured using a CMOS process that allows high level of integration to design a dedicated circuit which is factory trimmed to better match the sensing element characteristics [20].

The ADXL345 has a user selectable full scale of ±2g, ±4g, ±8g, ±16g and it is capable of measuring acceleration about a bandwidth of 1600 Hz for all axes. The device bandwidth may be selected accordingly to the application requirements. The self-test capability allows the user to check the functioning of the device. The ADXL345 is available in plastic SMD package and it is specified over a temperature range extending from -40°C to +85°C [20].

The data logger shown in Figure 3.7 is manufactured for the ADXL345 tests by ADI and we collected the most of the data in the thesis to make practice time and again. We collected the data directly to the PC to develop the accurate software easily.

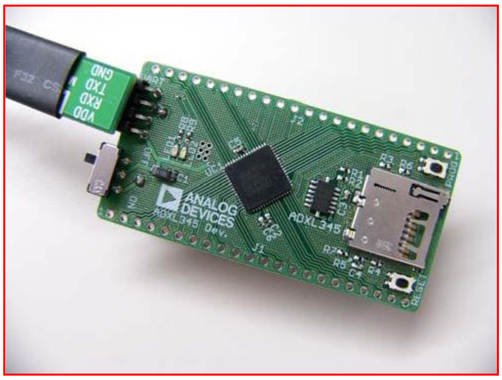

Figure 3.7: ACG Data Logger (DL)

The power supply of the SN is directly applied to the circuit. The battery used in the SN is a 1/2 AA Lithium battery. This battery and/or energy harvesting charger could be choosen at the different capacities and the types for the implant application. According to the measurements, the average current consumption of the SN is about 25 mA during bursts and under 100 uA during normal operation. Hence, the total current consumption of the board is still around 100 micro-A because the system does not send any data if not requested or if there is not any cardio rhythm abnormality.

3.3 The Sensor Access Point (SAP)

The SAP is based on the same technology with the sensor node (SN). SAP has additional USB interface, a test screen and a mains power supply input option. The SAP sends or receives information to or from the SN. In the thesis it doesn't have any other task than being an access point for the data comes from the SN. However, it could be used for the optional following purposes:

- To collect the data generated by more than one sensor node.
- SAP could be used to process the ACG data directly and communicate with the remote entities (i.e., the mobile users, PC in a hospital etc.) via GPRS modem or ADSL modem etc. without a PC [19].
- Once the data collected by the SAP, it could be stored in a database or file system with a certain format, or it could be visualized and displayed in a graphical user interface (GUI) or displayed as web content as a stand alone medical device.

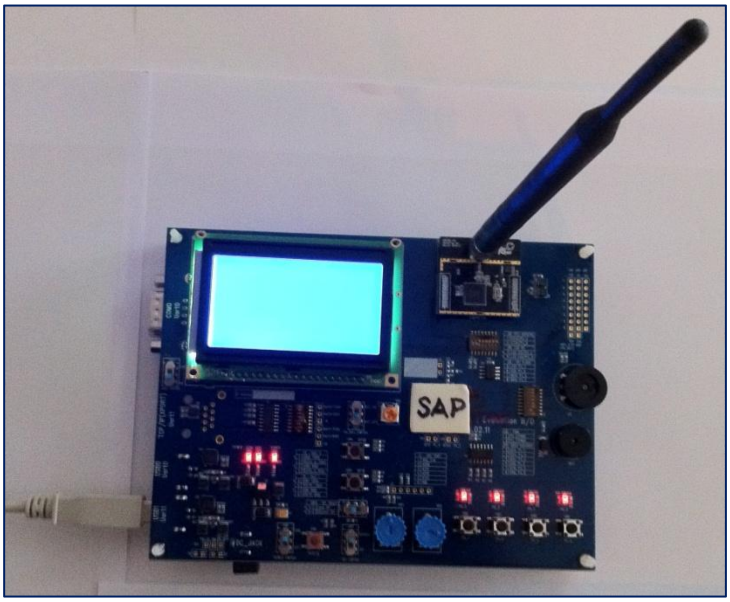

Figure 3.8: Sensor Access Point (SAP).

The SAP doesn't make any process on the data and transmits the ACG data to the PC without making any change and the PC is going to process the data with the cardio rhythm analysis software and display it on the browser furthermore the PC will serve it on the web to be monitored by any mobile device.

3.4 The Application Software and the PC Server

An embedded, multi-threaded, C-based application program has been developed for the Personal Logger and Sensor Node by Keil IDE. And we developped a server program to analize the collected data with the PHP. The PHP is a platform independent and browser based programming language. We uploaded the program to a server for the experiment of the thesis. Therefore we can check our experiment results over any mobile phone with the browser and/or PC [6].

4. THE PROCESSES OF THE CARDIO RHYTHM ANALYSIS SYSTEM

The system has the following parts to realize the monitoring of the cardio rhythm:

- Data Acquisition: The previously collected heartbeat data, i.e. acceleration data, will be processed to show the time differences between the heartbeat pulses.
- Module Communication: To send/receive information with the IEEE802.15.4 standard devices which are the most innovative “Wireless Body Area Network” (WBAN) technology. The data communication with a basic WBAN will be demonstrated.
- Computer/Mobile Device Interfacing: The data processing application will be demonstrated on the PC and on the internet with the software developed on the PHP to show a browser based platform independent application which can be used with any smart phone or PC [6].

In the thesis, following the collection of the data with the accelerometer, the data were transfered to a PC to be analyzed by the zero-crossing algorithm and then served on the internet. The algoritm is a modified version of the zero-crossing algorithm. Because, the zero-crossing algorithm described in the litriture was for the ECG signals. Since the shape and nature of the ACG signal is very much different than the ECG signal the original algorithm has been changed.

The analysis method is based on a feature obtained by counting the number of zero-crossings per segment. It is well-known that zero-crossing methods are robust against noise and are particularly useful for finite precision arithmetic. The new detection method inherits this robustness and provides a high degree of detection performance even in cases of very noisy electrocardiographic signals. Furthermore, due to the simplicity of detecting and counting zero-crossings, the proposed technique provides a computationally efficient solution to the heart beat rate detection problem [14].

4.1 The Flow of the ACG Data Processing Algorithm with Zero-Crossing

Algorithm Overview: Figure 4.1 shows the block diagram of the algorithm. The algorithm used in this thesis consists of six stages: the data collection, the extraction of the feature signal, the event detection, and the temporal localization of the heart beat, the detection and the result sending, the sensitivity calculation

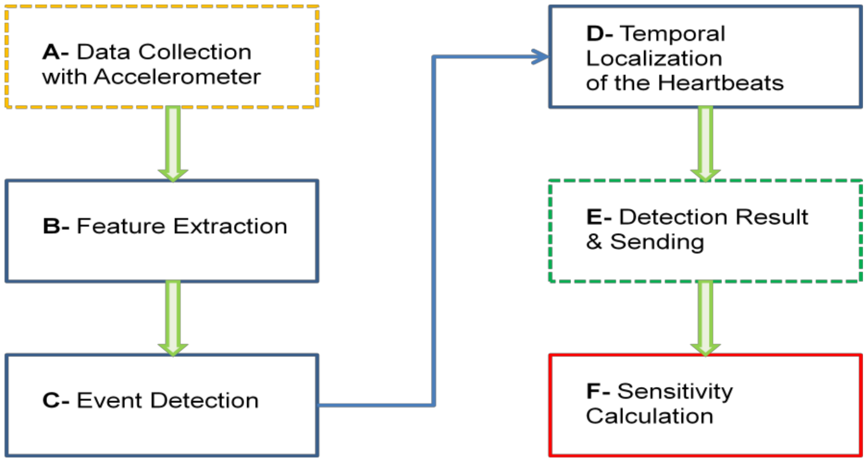

Figure 4.1: The stages of the rhythm analysis algorithm.

4.2 Data Collection

The data sample collection rate is 100 Hz thus the band width of the signal to be analized will be 50 Hz maximum and it is quite good to apply the proposed algorithm to the accelerometer heartbeat data [13]. Since the accelerometer has oscillated data nature we shall keep the whole data at the positive side of the time axis. We shall decrease the DC content from the main data colletion. We apply the non-linear transformation to make the heart-beat information much clear. We add the frequency component to make the zero crossings.

As in most conventional algorithms, bandpass filtering is performed as a preprocessing method in order to increase the signal to noise ratio, i.e., to attenuate the mean as well as some high-frequency noise.

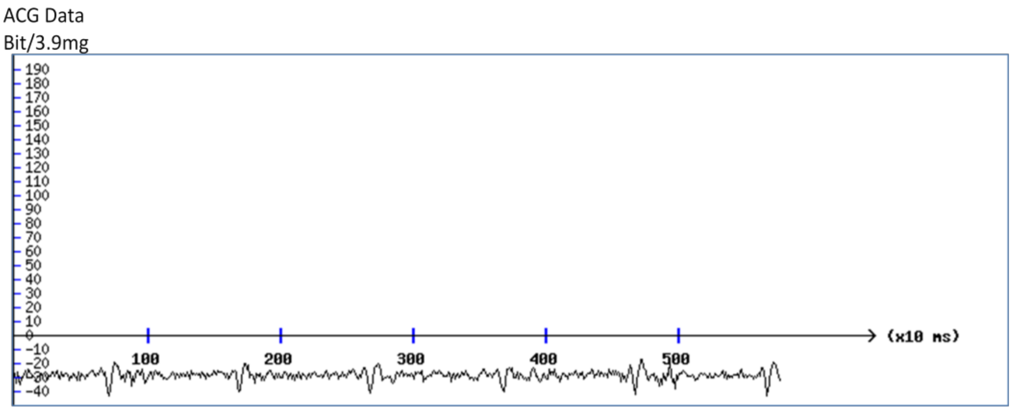

Figure 4.2: The collected raw ACG data by the accelerometer.

4.3 Feature extraction

The frequency content of a hearbeat data for the HS may extend up to 40 Hz to make a reliable analysis and ACG has been accepted in the same frequency limits in the thesis. Since we have 100 Hz sampling rate and thus 50 Hz band-width it is not necessary to apply low pass filter. We can describe the feature extraction process by using the Figure 4.3 below:

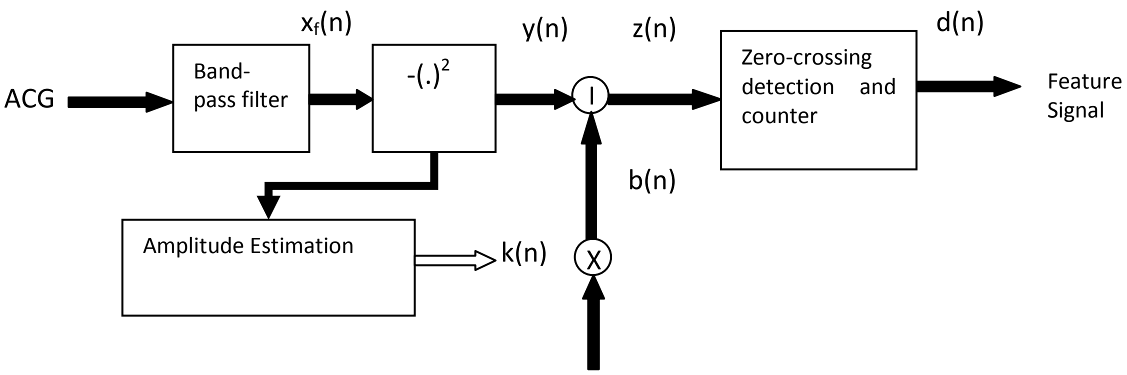

Figure 4.3: Block diagram of the feature extraction.

The block diagram of the ACG feature extraction algorithm is shown in Figure 4.6. It consists of a band-pass filter, a non-linear transform, the amplitude estimation and substarction of the high-frequency sequence, the zero crossing detection, and the zero crossing comparison. After the comparison we will determine a value N, e.g. N value is fifty in the thesis, for the positive results and zero for the zero or the negative results. The input signal is the original ACG signal, and the output is the feature signal which shows the G1 complex location.

The cut-off frequencies are 0 Hz and 50 Hz. In this case a simple filter which eleminates the DC component of the ACG sisgnal is sufficient. This is one of the advantages of the ACG analysis method with the accelerometer sensor. Due to the insignificant highfrequency components, the low pass part of the bandpass had been omitted. This is another deduction of the calculation overhead.

Due to the spectral characteristics of the ACG components, it is reasonable to filter the ACG signal at first in order to attenuate the mean. Because the filtered signal will be used for the temporal localization of the heart beat reference point, it is important to use a bandpass filter with a linear phase response. Otherwise, an accurate localization of the heart beat would be impossible.

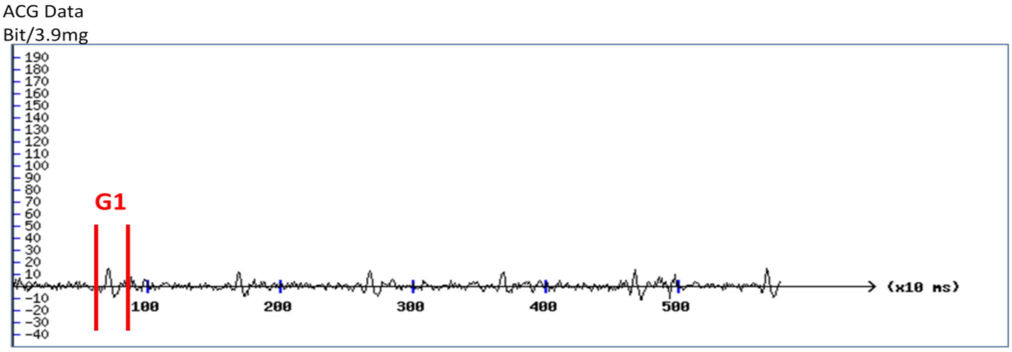

Figure 4.4: Bandpass filtered ACG data $x_f(n)$ oscillating around zero and the G1 complex.

As depicted in Figure 4.4 for the ACG signals, the bandpass filtered signal oscillates around zero. If it is though that the ACG signal propagates like the HS signal we may assume that during the S1 complex of the HS, the G1 complex of the ACG can be observed. The G1 complex has the high amplitude because of the high pressure waves sensed by the ACG sensor; otherwise the signal amplitude would be low.

A further increase in the signal quality is achieved by a non-linear transform of the signal as shown in Figure 4.5:

$$y(n) = \mathrm{sign}(x_f(n)) \,.\, x_f(n) \qquad \textbf{(4.1)}$$

where x_f (n) denotes the bandpass filtered ACG and y(n) denotes the non-linearly transformed $x_f(n)$ signal. The signal y(n) will be used later for the determination of the event and the temporal location of the G1-wave.

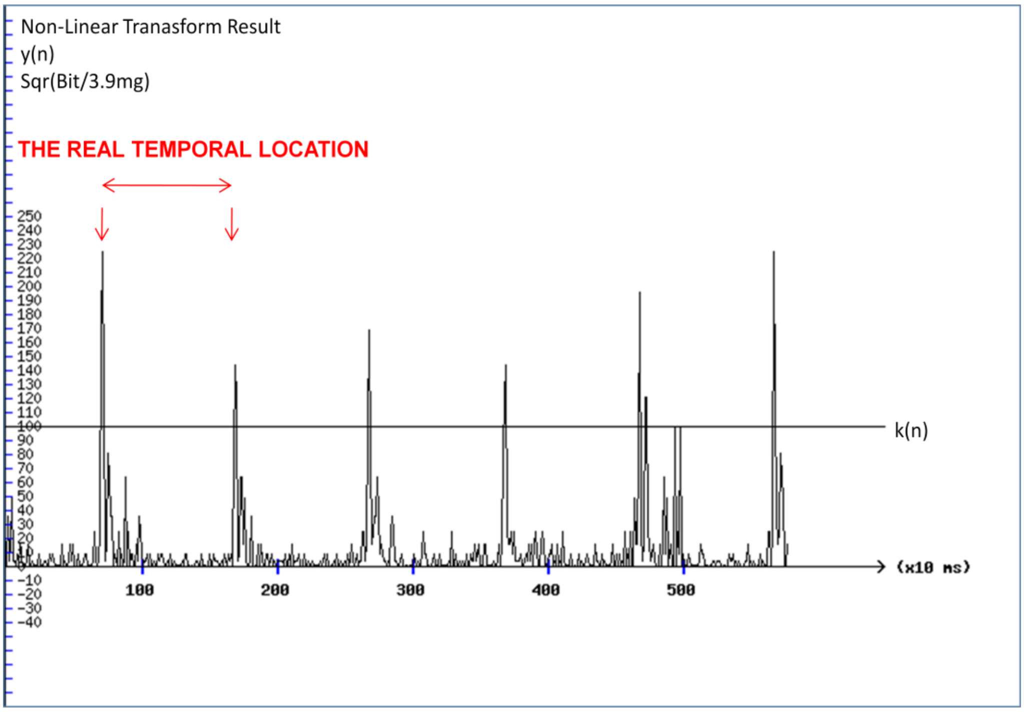

Figure 4.5: Non-linear transformation of the bandpass filtered ACG signal y(n).

After the non-linear transformation output it is necessary to add a negative-high-frequency sequence in order to increase the number of zero crossings during non-G1 segments.

$$b(n) = (-1) . k(n) \quad \textbf{(4.2)}$$

to the signal, i.e.,

$$z(n) = y(n) + b(n) \quad \textbf{(4.3)}$$

In the ideal case, the feature signal d(n) assumes the value d(n) = 0 during non-G1 segments and d(n) = N during the G1 complex as depicted in Figure 4.6. An arbitrary N value is 50 as we declared before.

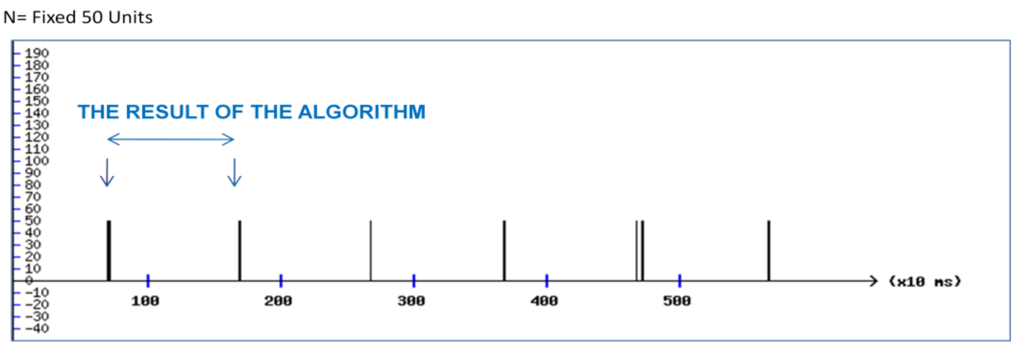

Figure 4.6: Feature signal d(n).

However, if the amplitude k(n) of the high-frequency sequence b(n) is too large, the number of zero crossings is always 0 (zero). If k(n) is too small, the feature signal is noisy, and the difference between the number of zero crossings during the G1 segments and non-G1 segments is not significant enough for a good classification. Hence, the amplitude k(n) of the sequence must be properly determined. In our implementation, k(n) is determined from the magnitude of the signal y(n) where $\lambda_K \in (0;1)$ is a balancing factor, e.g. $\lambda_K=1$ and the design parameter c denotes a constant gain, e.g., c = 100.

$$\mathrm{k(n)} = \lambda_K \, . \, \mathrm{c} \tag{4.4}$$

There are several methods for the detection of zero-crossings; for an overview see [14]. In our implementation we use the following comparison equation:

$$d(n) = \begin{cases} 0, & z(n) \geq 0 \\ N, & z(n) < 0 \end{cases} \tag{4.5}$$

Then, the number of zero-crossings per segment is usually computed by:

$$\frac{1}{N}\sum_{i=0}^{n-1} d(n) \tag{4.6}$$

Characteristically the feature signal assumes high values during a G1 complex and low values otherwise. The maximum value of the feature signal is given by the length W of the moving window. The number of zero-crossing per segment is an equivalent representation of the frequency of the non-G1 component of a signal segment. It can be considered as roughly proportional to the non-dominant frequency. Accordingly, for a low-frequency and high-amplitude component one can expect fewer zero crossings per segment than for a high-frequency and low-amplitude component. After the addition of the negative-high-frequency sequence b(n) and the comparison algorithm (4.5) with N value, the signal is dominated by a low-frequency and high-amlitude oscillations during the G1 complex and dominated by the positive-high-frequency sequence, it means zero otherwise. Hence, the number of zero-crossings must be low during the G1 complex and high at all other times. After the comparison algorithm with N =50 we see the feature signal result as in Figure 4.6.

4.4 Event Detection

An event begins when the feature signal (a number of N valued samples per segment) rises above adaptive threshold as shown in Figure 4.6. The events ends when the signal falls below the threshold. Both the beginning and the end of the event are the boundaries of the search interval for the temporal localization of the G1 complex. If adjacent events are temporally very close (multiple events), they will be combined into one single event. The beginning of the combined event is the beginning of the first event, and the end of the combined event is the end of the last event. This combined event is the real event for the temporal location analysis of the G1 complex. This serach is done in the boundaries of the W moving window intervals. W=10 in the example, it means the algorithm is counting 10 sample location to make the comparison for the event detection calculation.

For the event detection, the distance between events is considered. This calculation is made in the moving window (W) borders. The event detection algorithm counts the zeros between events and if the zeros are equal or less than five the events are combined and considered as the one single event, otherwise it is considered as an adjacent independent event by the event detection algorithm.

4.5 Temporal Localization of the G1 Complex

The detection of the G1 complex is completed by the determination of the temporal location of the G1 complex. The temporal location is determined by a single event or combined-single event first sample location search.

The beginning and the end of the event, i.e., the temporal location of the event, provide the bounds for the search interval that is used for the temporal localization of the G1 complex to count the heartbeats. As depicted in Figure 4.6, the beginning of each single event is considered as the temporal location of the event. Thus the algorithm calculates the heart beats and the time intervals between the beats by making the beginning point analysis.

4.6 Detection Results and Sending with Sensitivity Calculation

The complete cardio rhythm analysis program had been attached as the Appendix A and the ACG data of a subject which was used in the example of the thesis had been attached as the Appendix B. The program and the data can be used to make further academical study.

In the example the detection results served by the PC on the web site. Any authorized user, e.g. doctors, can connect to the web site and can see the results. The same cardio rhythm analysis algorithm could be embedded into the sensor node (SN) MCU thus the calculations would be done in the sensor node (SN) by the same algorithm and it would be sent to any device directly from the patient.

The results and the sensitivity will be discussed in the last section as the success of the system and the analysis method.

5. CONCLUSION AND DISCUSSIONS

To determine the success of the cardio rhythm analysis system, the sensitivity will be calculated by using the ACG data set at the Appendix B. In the data set there are six temporal location measurements of a subject. Then the calculations will be extended to four subjects' data to reach more reliable sensitivity result.

5.1 Sensitivity Calculation

The sensitivity describes the percentage of the closeness of the calculated signal temporal location to the real signal temporal location. This is also called as the positive sensitivity. If the ratio of the error to the real data was calculated, it would be named as the negative sensitivity. To find the sensitivity, the method is to compare the error of the calculated signal temporal location with real signal temporal location. This will be the negative sensitivity. The substraction of the negative sensitivity from the value 100 will translate the result as the positive sensitivity. The positive sensitivity will be called as the sensitivity.

The calculation result and the captured data measuring the captured data form the plotting on a sheet.

$$Sensitivity(m) = 100 - \frac{The\ Error\ of\ the\ Result\ Signal\ Temporal\ Location(m)}{The\ Real\ Signal\ Temporal\ Location(m)} \qquad \textbf{(5.1)}$$

The following Table 5.1 will show the caculations of the real timings and the measurement errors with the ACG algorithm. The temporal locations of the real G1 complex and the calculated version with the algorithm can be seen in Figure 4.5 and Figure 4.6. Max value square value was used to find the exact location of the real signal. In the table, m is the number of the G1 complexes.

Table 5.1: The real and calculated time locations of the six G1 complexes.

Index (m)	***Calculated Location***	***Real Location***	***Max Value Square***
1	71	72	144
2	169	170	225
3	268	269	169
4	369	370	144
5	468	469	196
6	567	568	225

By using Table 5.1 the following calculation could be done to determine the heartbeat times. By using the difference of the real temporal locations of the G1 complex and the calculated temporal location of the G1 complex.

$$Real\_Beat\_Time(m) = Real\_Loc(m) - Real\_Loc(m-1) \tag{5.2}$$

$$Calculated\_Beat\_Time(m) = Calc\_Loc(m) - Calc\_Loc(m-1) \tag{5.3}$$

$$Error(m) = Real\_Beat\_Time(m) - Calculated\_Beat\_Time(m) \tag{5.3}$$

Table 5.2: The real and calculated heartbeat times of the six G1 complexes.

Index (m)	***Real Beat Time (10ms)***	***Calculated Beat Time (10ms)***	***Error***
1	-	-	-
2	98	98	0
3	99	99	0
4	101	101	0
5	99	99	0
6	99	99	0

By using the equation (5.1), the sensitivity of each G1 complex measurement of the algorithm can be done. However, to find the overall sensitivity of the cardio rhythm analysis system, the avarage sensitivity calculation should be done. If M is the number of the G1 complexes:

$$Sensitivity(M) = \frac{1}{M-1} \sum_{m=2}^{M} Sensitivity(m) \quad \textbf{(5.4)}$$

By using the equation (5.4) the sensitivity results of the Table 5.2 contents can easily be calculated. Due to the initial reference measurement need, the first measurement's relative temporal location can not be calculated. It means m=1 will be an initial reference point however m=1 itself will not be taken into consideration in the sensitivity calculation. Therefore for the six G1 complexes, M will be 6 and the calculation result of the Sensitivity(M) will be as follows:

$$Sensitivity(6) = \frac{1}{6-1} \sum_{m=2}^{6} Sensitivity(m) \quad \textbf{(5.6)}$$

The result of the above mentioned equation by using Table 5.2 would be as follows:

$$Sensitivity(6) = \%100 \quad \textbf{(5.6)}$$

If the sensitivity analysis is extended to the higher number (e.g. Z value) of G1 complexes from the different subjects the sensitivity value will be probabilisticaly more closer to the real sensitivity of the proposed algorithm. Therefore we extended the experiments to the four different subjects with five G1 complexes sampled from each. We naturally used different balancing factors λ_K and we fount the sensitivity as follows:

$$Sensitivity(Z) = \frac{1}{Z} \sum_{k=1}^{Z} Sensitivity(M_k) \quad \textbf{(5.7)}$$

Above mentioned equation is depicting that each discrete group of G1 complex collection session is calculated using $Sensitivity(M_k)$ formula as shown in the equation (5.4). Then Z different measurement i.e. different subjects here are calculated and avaraged. $M_k$is the sample number for the each different session. The reason of this discrete calculation is the missing initial G1 complex for each calculation. Thus, our result by using formula (5.7) is

$$Sensitivity(5) = \frac{1}{5} \sum_{k=1}^{5} Sensitivity(M_k) \quad \textbf{(5.8)}$$

$$Sensitivity(5) = \%99 \quad \textbf{(5.9)}$$

5.2 Discussions

If physiologically reasonable forgetting factors and timeout periods are chosen, the algorithm works reliably, also in cases of minor parameter maladjustments. Due to the non-linear type of the algorithm, the impact of the different parameters on the performance is not independent from each other. However, from our experience with this algorithm, the impact of these interdependencies does not lead to real tuning problems. The band pass filter has a great impact on the performance of the algorithm [14]. It is most important that this filter removes the non-G1 waves. Hence, this filter needs to be chosen carefully to reach to the higher performances.

For derivative-based feature signals, variations in the amplitude level or the morphology of the R peak result in changes of derivative based feature signals [1]. In contrast to this, the number of zero crossings does not change with the amplitude variations in problematic sections of the signal. This characteristic leads to a significant improvement of the detection performance on the level of computational expensive algorithms.

Of course, the sensitivity depends on the amplitude k(n) of the highfrequency sequence b(n). However, this amplitude is easily controllable by the algorithm. The algorithm inherits its robustness against noise from the general robustness of zero crossing signal processing methods against noise. The combination of temporarily close events into one single event replaces the usual refractory period. Furthermore, it prevents the execution of several R-wave searches for only one QRS complex as in the common ECG algorithms and hence increases the efficiency of the algorithm. The computational load is low. The computationally most consuming stages are the band pass filter and the maximum/minimum search for the temporal localization of the G1 complex.

The algorithm is making a morphological analysis therefore it is very suitable for the ACG analysis which was performed in the SN. The performance of the algorithm is comparable to other algorithms reported in the literature. Due to this simple principle, G1-G1 detection smilar to R-R detection can be realized at low computational costs. The algorithm catches all of the G1 complex and the general negative sensitivity i.e. error rate of 1% is coming from the miscombining of the G1 complexes during the temporal localization part of the algorithm. It can be easily improved by developing better bandpass filtering and more accurate balancing factor but the most important part is to keep the power consumption as low as possible to be able to use the system as an implant in the body with the energy harvesters.

5.3 Future Expectations

A prototype of wireless cardio rhythm monitoring with the accelerometer sensor was developed for the future homecare or personalcare needs as well as patient monitoring in the hospitals or any medical aid suppliers. Our system acts also as a continuous event recorder where the patient is unconscious and lying on the bed. The use of an affordable device for monitoring cardio activities and analyzing cardio rhythm can provide informative details to the doctors in a ubiquitious way, and simultaneously alert the doctor of any emergencies. The goal is to provide a capability for real time monitoring of any chronical cardio rhythm based decease at home using accelerometer sensors and then notify to the doctor's PC or PDA [5].

In the next step; with the other accelerometers implanted into the different locations of the body, especially by fixing onto the certain places in the body on the bone very close to heart vibrations, we may acquire an excellent filtering of moving patient by using more efficient digital filtering methods as well [24].

To highlight the importance of such systems we may declare the number of the people who dies from heart attack is 18 million in the world, 300,000 in USA and 200,000 in Turkey. It is expected to be 25 million in 2020. with the 40% increase. This figure is 1.5 million for the taffic accidents [22]. The number of the potential heart decease can be guessed with the number of the 33.82% potential metabolic syndrome risk in our country according to the periodic researches of METSAR - Metabolic Syndrome Research Group. It’s very clearly seen that It is inevitable to make researches to find practical solutions to the heart diseases for the health of human being.

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APPENDICES

APPENDIX A.1 : The PHP program to draw the Subject's data.
APPENDIX A.2 : The PHP program to draw the band-pass filter output.
APPENDIX A.3 : The PHP program to draw the non-linear transformed data.
APPENDIX A.4 : The PHP program to make the zero-crossing analysis.
APPENDIX B : The data of the Subject used in the examples of the thesis.

APPENDIX A.1

```
<?php
$x = 750;
$y = 250;
$gd = imagecreatetruecolor($x, $y);
$red = imagecolorallocate($gd, 0, 0, 0);
imagefill($gd , 0,0 , imagecolorallocate($gd, 255,255,255));
$fcontents = file('ThesisSubject_DATA001.TXT');
$toplam = 0;
$a = 0;
for($i=1; $i<sizeof($fcontents); $i++) {
	$line = trim($fcontents[$i]);
	$arr = explode("\t", $line);
	$x = str_replace(',',",$arr[3]);
	$toplam = $toplam + $x;
	$a++;
}
$toplam = round($toplam / ($a + 1));
$a = 0;
$yatay = 0;
$dikey = 10;
for($i=1; $i<sizeof($fcontents); $i++) {
	$line = trim($fcontents[$i]);
	$arr = explode("\t", $line);
	$x = str_replace(',',",$arr[3]);
	imagesetpixel($gd, 0,$dikey, $red);
	imageline($gd,0,$dikey,5,$dikey,255);
	if($yatay==100){
		imagesetpixel($gd, $i,195, $red);
		imageline($gd,$i,195,$i,205,255);
		imageline($gd,$i+1,195,$i+1,205,255);
		$yatay = 0;
	}
	$x = 200 - $x;
	imagesetpixel($gd, $a,$x, $red);
	imageline($gd,$a,$x,$old_salise,$old_x,0);
	$old_salise = $a;
	$old_x = $x;
	$a++;
	$yatay++;
	$dikey = $dikey + 10;
}
imagestring($gd, 3, 645, 194,  "> (x10 ms)", $red);
$zz = 233;
$yy = -40;
for($z=0;$z<=23;$z++){
	imagestring($gd, 2, 10, $zz, $yy, $red);
	$zz = $zz - 10;
	$yy = $yy + 10;
}
imagestring($gd, 3, 90, 210,  "100", $red);
imagestring($gd, 3, 190, 210,  "200", $red);
imagestring($gd, 3, 290, 210,  "300", $red);
imagestring($gd, 3, 390, 210,  "400", $red);
imagestring($gd, 3, 490, 210,  "500", $red);
imageline($gd,0,0,0,1400,0);
imageline($gd,0,200,650,200,0);
header('Content-Type: image/png');
imagepng($gd);
?>
```

APPENDIX A.2

```
<?php
$x = 750;
$y = 250;
$gd = imagecreatetruecolor($x, $y);
$red = imagecolorallocate($gd, 0, 0, 0);
imagefill($gd , 0,0 , imagecolorallocate($gd, 255,255,255));
$fcontents = file('ThesisSubject_DATA001.TXT');
$toplam = 0;
$a = 0;
for($i=1; $i<sizeof($fcontents); $i++) {
	$line = trim($fcontents[$i]);
	$arr = explode("\t", $line);
	$x = str_replace(',','',$arr[3]);
	$toplam = $toplam + $x;
	$a++;
$toplam = round($toplam / ($a + 1));
$a = 0;
$yatay = 0;
$dikey = 10;
for($i=1; $i<sizeof($fcontents); $i++) {
	$line = trim($fcontents[$i]);
	$arr = explode("\t", $line);
	$x = str_replace(',','',$arr[3]);
	if(str_replace('-','',$x)){
		$x = 200 + $x;
	} else {
		$x = 200 - $x;
	}
	$x = $x - $toplam;
	imagesetpixel($gd, 0,$dikey, $red);
	imageline($gd,0,$dikey,5,$dikey,255);
	if($yatay==100){
		imagesetpixel($gd, $i,195, $red);
		imageline($gd,$i,195,$i,205,255);
		imageline($gd,$i+1,195,$i+1,205,255);
		$yatay = 0;
	}
	imagesetpixel($gd, $a,$x, $red);
	imageline($gd,$a,$x,$old_salise,$old_x,0);
	$old_salise = $a;
	$old_x = $x;
	$a++;
	$yatay++;
	$dikey = $dikey + 10;
}
imagestring($gd, 3, 645, 194,  "> (x10 ms)", $red);
$zz = 233;
$yy = -40;
for($z=0;$z<=23;$z++){
	imagestring($gd, 2, 10, $zz, $yy, $red);
	$zz = $zz - 10;
	$yy = $yy + 10;
}
imagestring($gd, 3, 90, 210,  "100", $red);
imagestring($gd, 3, 190, 210,  "200", $red);
imagestring($gd, 3, 290, 210,  "300", $red);
imagestring($gd, 3, 390, 210,  "400", $red);
imagestring($gd, 3, 490, 210,  "500", $red);
imageline($gd,0,0,0,1400,0);
imageline($gd,0,200,650,200,0);
header('Content-Type: image/png');
imagepng($gd);
?>
```

APPENDIX A.3

```
<?php
$x = 750;
$y = 500;
$gd = imagecreatetruecolor($x, $y);
$red = imagecolorallocate($gd, 0, 0, 0);
imagefill($gd , 0,0 , imagecolorallocate($gd, 255,255,255));
$fcontents = file('ThesisSubject_DATA001.TXT');
$toplam = 0;
$a = 0;
for($i=1; $i<sizeof($fcontents); $i++) {
	$line = trim($fcontents[$i]);
	$arr = explode("\t", $line);
	$x = str_replace(',','',$arr[3]);
	$toplam = $toplam + $x;
	$a++;
}
$toplam = round($toplam / ($a + 1));
$a = 0;
$yatay = 0;
$dikey = 10;
for($i=1; $i<sizeof($fcontents); $i++) {
	$line = trim($fcontents[$i]);
	$arr = explode("\t", $line);
	$x = str_replace(',','',$arr[3]);
	$x = $x - $toplam;
	$x = $x * $x;
	imagesetpixel($gd, 0,$dikey, $red);
	imageline($gd,0,$dikey,5,$dikey,255);
	if($yatay==100){
		imageline($gd,$i,405,$i,395,255);
		imageline($gd,$i+1,405,$i+1,395,255);
		$yatay = 0;
	}
	$x = 400 - $x;
	imagesetpixel($gd, $a,$x, $red);
	imageline($gd,$a,$x,$old_salise,$old_x,0);
	$old_salise = $a;
	$old_x = $x;
	$a++;
	$yatay++;
	$dikey = $dikey + 10;
}
imagestring($gd, 3, 645, 394,  "> (x10 ms)", $red);
$zz = 433;
$yy = -40;
for($z=0;$z<=29;$z++){
	imagestring($gd, 2, 10, $zz, $yy, $red);
	$zz = $zz - 10;
	$yy = $yy + 10;
}
imagestring($gd, 3, 90, 410,  "100", $red);
imagestring($gd, 3, 190, 410,  "200", $red);
imagestring($gd, 3, 290, 410,  "300", $red);
imagestring($gd, 3, 390, 410,  "400", $red);
imagestring($gd, 3, 490, 410,  "500", $red);
imageline($gd,0,0,0,1400,0);
imageline($gd,0,400,650,400,0);
imageline($gd,0,300,650,300,0);
header('Content-Type: image/png');
imagepng($gd);
?>
```

APPENDIX A.4

```
<?php
$x = 750;
$y = 250;
$gd = imagecreatetruecolor($x, $y);
$red = imagecolorallocate($gd, 0, 0, 0);
imagefill($gd , 0,0 , imagecolorallocate($gd, 255,255,255));
$fcontents = file('ThesisSubject_DATA001.TXT');
$toplam = 0;
$a = 0;
for($i=1; $i<sizeof($fcontents); $i++) {
        $line = trim($fcontents[$i]);
        $arr = explode("\t", $line);
        $x = str_replace(',',",$arr[3]);
        $toplam = $toplam + $x;
        $a++;}
$toplam = round($toplam / ($a + 1));
$a = 0;
$yatay = 0;
$dikey = 10;
for($i=1; $i<sizeof($fcontents); $i++) {
        $line = trim($fcontents[$i]);
        $arr = explode("\t", $line);
        $x = str_replace(',',",$arr[3]);
        $x = $x - $toplam;
        $x = $x * $x;
        $x = $x - 100;
        if($x<=0){
                $x = 0;
        } else {
                $x = 50;
        }
        if($x==50){
                imagesetpixel($gd, $a,150, $red);
                imageline($gd,$a,150,$a,200,0);}
        imagesetpixel($gd, 0,$dikey, $red);
        imageline($gd,0,$dikey,5,$dikey,255);
        if($yatay==100){
                imagesetpixel($gd, $i,195, $red);
                imageline($gd,$i,195,$i,205,255);
                imageline($gd,$i+1,195,$i+1,205,255);
                $yatay = 0;
        }
        $old_salise = $a;
        $old_x = $x;
        $a++;
        $yatay++;
        $dikey = $dikey + 10;
}
imagestring($gd, 3, 645, 194,  "> (x10 ms)", $red);
$zz = 233;
$yy = -40;
for($z=0;$z<=23;$z++){
        imagestring($gd, 2, 10, $zz, $yy, $red);
        $zz = $zz - 10;
        $yy = $yy + 10;}
imagestring($gd, 3, 90, 210,  "100", $red);
imagestring($gd, 3, 190, 210,  "200", $red);
imagestring($gd, 3, 290, 210,  "300", $red);
imagestring($gd, 3, 390, 210,  "400", $red);
imagestring($gd, 3, 490, 210,  "500", $red);
imageline($gd,0,0,0,1400,0);
imageline($gd,0,200,650,200,0);
header('Content-Type: image/png');
imagepng($gd);
?>
```

APPENDIX B

ndx:	HR:MN:SC:1/128s:	x:	y:	z:	ndx:	HR:MN:SC:1/128s:	x:	y:	z:
137	00:00:06:015	-244	-26	62	**184**	00:00:06:019	-242	-30	58
138	00:00:06:015	-242	-28	71	**185**	00:00:06:019	-241	-32	59
139	00:00:06:015	-235	-34	65	**186**	00:00:06:019	-247	-26	56
140	00:00:06:015	-235	-26	65	**187**	00:00:06:019	-242	-32	53
141	00:00:06:015	-238	-32	61	**188**	00:00:06:019	-242	-27	58
142	00:00:06:016	-247	-35	61	**189**	00:00:06:019	-242	-28	49
143	00:00:06:016	-248	-30	55	**190**	00:00:06:020	-242	-29	56
144	00:00:06:016	-251	-30	52	**191**	00:00:06:020	-242	-31	56
145	00:00:06:016	-251	-28	50	**192**	00:00:06:020	-244	-28	58
146	00:00:06:016	-251	-25	46	**193**	00:00:06:020	-245	-29	49
147	00:00:06:016	-245	-29	49	**194**	00:00:06:020	-246	-27	56
148	00:00:06:016	-246	-24	52	**195**	00:00:06:020	-245	-26	57
149	00:00:06:016	-245	-26	55	**196**	00:00:06:020	-246	-25	57
150	00:00:06:016	-244	-27	55	**197**	00:00:06:020	-245	-25	58
151	00:00:06:016	-241	-29	60	**198**	00:00:06:020	-242	-28	58
152	00:00:06:016	-243	-29	60	**199**	00:00:06:021	-245	-29	57
153	00:00:06:017	-242	-30	56	**0**	00:00:07:001	-247	-27	61
154	00:00:06:017	-245	-32	59	**1**	00:00:07:002	-248	-28	58
155	00:00:06:017	-245	-29	55	**2**	00:00:07:002	-250	-26	58
156	00:00:06:017	-242	-29	57	**3**	00:00:07:002	-245	-23	58
157	00:00:06:017	-245	-30	54	**4**	00:00:07:002	-236	-26	63
158	00:00:06:017	-246	-27	52	**5**	00:00:07:002	-232	-28	61
159	00:00:06:017	-246	-29	53	**6**	00:00:07:002	-234	-26	57
160	00:00:06:017	-246	-27	50	**7**	00:00:07:002	-239	-33	52
161	00:00:06:017	-246	-29	51	**8**	00:00:07:002	-245	-41	47
162	00:00:06:017	-243	-31	54	**9**	00:00:07:002	-251	-43	47
163	00:00:06:017	-247	-29	54	**10**	00:00:07:002	-253	-39	47
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165	00:00:06:017	-244	-29	58	**12**	00:00:07:002	-251	-24	42
166	00:00:06:017	-246	-26	56	**13**	00:00:07:002	-251	-19	51
167	00:00:06:017	-243	-26	64	**14**	00:00:07:002	-251	-20	57
168	00:00:06:018	-242	-29	64	**15**	00:00:07:003	-245	-22	62
169	00:00:06:018	-245	-27	60	**16**	00:00:07:003	-246	-22	63
170	00:00:06:018	-245	-25	61	**17**	00:00:07:003	-238	-27	67
171	00:00:06:018	-243	-30	53	**18**	00:00:07:003	-235	-31	66
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177	00:00:06:018	-247	-27	50	**24**	00:00:07:003	-247	-27	39
178	00:00:06:018	-243	-28	55	**25**	00:00:07:003	-252	-26	36

179	00:00:06:018	-242	-32	53
180	00:00:06:018	-244	-29	56
181	00:00:06:018	-244	-30	57
182	00:00:06:018	-244	-27	56
183	00:00:06:019	-240	-29	54
31	00:00:07:004	-245	-25	57
32	00:00:07:004	-245	-28	64
33	00:00:07:004	-248	-28	66
34	00:00:07:005	-241	-24	67
35	00:00:07:005	-236	-30	75
36	00:00:07:005	-232	-34	61
37	00:00:07:005	-235	-33	62
38	00:00:07:005	-242	-31	59
39	00:00:07:005	-246	-29	57
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27	00:00:07:003	-247	-29	46
28	00:00:07:004	-246	-33	44
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79	00:00:07:008	-244	-26	60
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88	00:00:07:009	-244	-25	58
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93	00:00:07:009	-246	-26	58
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97	00:00:07:010	-242	-30	60
98	00:00:07:010	-248	-27	55
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109	00:00:07:120	-252	-30	48
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111	00:00:07:121	-248	-20	46
112	00:00:07:121	-248	-20	54
113	00:00:07:121	-245	-24	60
114	00:00:07:121	-248	-21	65
115	00:00:07:121	-241	-30	69
116	00:00:07:121	-241	-28	68
117	00:00:07:121	-228	-28	70
118	00:00:07:121	-238	-30	73

72	00:00:07:007	-245	-28	59	**119**	00:00:07:121	-242	-34	58
73	00:00:07:007	-244	-27	58	**120**	00:00:07:121	-242	-27	57
74	00:00:07:007	-247	-28	55	**121**	00:00:07:121	-245	-28	50
75	00:00:07:007	-243	-27	59	**122**	00:00:07:122	-251	-29	49
76	00:00:07:008	-243	-28	58	**123**	00:00:07:122	-250	-27	44
77	00:00:07:008	-243	-27	58	**124**	00:00:07:122	-249	-32	41
125	00:00:07:122	-248	-29	38	**172**	00:00:07:126	-246	-26	55
126	00:00:07:122	-242	-32	46	**173**	00:00:07:127	-245	-25	53
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131	00:00:07:123	-244	-25	64	**178**	00:00:07:127	-241	-27	55
132	00:00:07:123	-242	-28	64	**179**	00:00:07:127	-246	-28	57
133	00:00:07:123	-242	-30	68	**180**	00:00:07:127	-248	-26	62
134	00:00:07:123	-237	-31	64	**181**	00:00:07:127	-244	-27	58
135	00:00:07:123	-239	-27	67	**182**	00:00:07:127	-246	-31	59
136	00:00:07:123	-238	-30	67	**183**	00:00:07:127	-246	-29	60
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138	00:00:07:123	-247	-30	57	**185**	00:00:07:127	-245	-26	53
139	00:00:07:123	-251	-30	46	**186**	00:00:07:127	-243	-29	52
140	00:00:07:123	-249	-30	53	**187**	00:00:07:127	-247	-28	57
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153	00:00:07:124	-243	-30	49	**0**	00:00:09:021	-242	-26	62
154	00:00:07:124	-247	-31	51	**1**	00:00:09:022	-238	-23	54
155	00:00:07:124	-247	-27	48	**2**	00:00:09:022	-232	-23	57
156	00:00:07:124	-247	-28	54	**3**	00:00:09:022	-233	-26	61
157	00:00:07:124	-242	-29	55	**4**	00:00:09:022	-241	-32	53
158	00:00:07:125	-244	-31	56	**5**	00:00:09:023	-248	-38	49
159	00:00:07:125	-241	-29	60	**6**	00:00:09:023	-251	-41	44
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161	00:00:07:125	-241	-30	59	**8**	00:00:09:023	-248	-32	47
162	00:00:07:125	-243	-27	60	**9**	00:00:09:023	-243	-23	51
163	00:00:07:125	-242	-27	64	**10**	00:00:09:023	-260	-22	51
164	00:00:07:125	-242	-26	60	**11**	00:00:09:023	-251	-22	62

165	00:00:07:125	-247	-28	61	**12**	00:00:09:023	-245	-20	60
166	00:00:07:125	-245	-29	58	**13**	00:00:09:023	-240	-22	70
167	00:00:07:125	-246	-29	53	**14**	00:00:09:023	-239	-25	71
168	00:00:07:125	-243	-29	56	**15**	00:00:09:023	-234	-32	67
169	00:00:07:126	-246	-28	53	**16**	00:00:09:023	-233	-31	66
170	00:00:07:126	-244	-30	58	**17**	00:00:09:023	-240	-30	66
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19	00:00:09:023	-243	-29	56	**66**	00:00:09:027	-246	-27	57
20	00:00:09:023	-247	-30	43	**67**	00:00:09:027	-247	-33	51
21	00:00:09:024	-251	-29	47	**68**	00:00:09:027	-246	-28	53
22	00:00:09:024	-246	-33	44	**69**	00:00:09:028	-245	-27	54
23	00:00:09:024	-248	-34	43	**70**	00:00:09:028	-244	-31	54
24	00:00:09:024	-246	-33	41	**71**	00:00:09:028	-244	-28	55
25	00:00:09:024	-246	-30	43	**72**	00:00:09:028	-245	-27	58
26	00:00:09:024	-246	-26	50	**73**	00:00:09:028	-245	-30	54
27	00:00:09:024	-246	-27	56	**74**	00:00:09:028	-243	-29	60
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29	00:00:09:024	-244	-26	64	**76**	00:00:09:028	-244	-28	60
30	00:00:09:024	-244	-25	65	**77**	00:00:09:028	-245	-28	60
31	00:00:09:024	-239	-29	69	**78**	00:00:09:028	-242	-25	59
32	00:00:09:024	-238	-28	65	**79**	00:00:09:028	-242	-28	52
33	00:00:09:024	-237	-27	68	**80**	00:00:09:028	-246	-28	57
34	00:00:09:024	-235	-29	64	**81**	00:00:09:028	-241	-31	56
35	00:00:09:024	-239	-28	58	**82**	00:00:09:028	-242	-30	57
36	00:00:09:024	-246	-29	56	**83**	00:00:09:028	-243	-30	53
37	00:00:09:025	-249	-27	54	**84**	00:00:09:028	-243	-32	54
38	00:00:09:025	-247	-28	49	**85**	00:00:09:029	-246	-30	48
39	00:00:09:025	-251	-31	46	**86**	00:00:09:029	-243	-31	53
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41	00:00:09:025	-244	-27	50	**88**	00:00:09:029	-247	-28	59
42	00:00:09:025	-242	-26	55	**89**	00:00:09:029	-246	-27	59
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44	00:00:09:025	-242	-27	64	**91**	00:00:09:029	-247	-24	58
45	00:00:09:025	-241	-25	61	**92**	00:00:09:029	-249	-24	58
46	00:00:09:026	-243	-33	64	**93**	00:00:09:029	-243	-27	52
47	00:00:09:026	-242	-31	66	**94**	00:00:09:030	-243	-29	58
48	00:00:09:026	-241	-31	56	**95**	00:00:09:030	-246	-29	53
49	00:00:09:026	-244	-29	56	**96**	00:00:09:030	-249	-27	57
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56	00:00:09:027	-244	-27	51	**103**	00:00:10:013	-235	-27	59
57	00:00:09:027	-244	-27	55	**104**	00:00:10:013	-241	-32	52

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64	00:00:09:027	-246	-29	61	**111**	00:00:10:013	-248	-23	54
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113	00:00:10:013	-242	-23	66	**160**	00:00:10:017	-241	-27	58
114	00:00:10:013	-236	-30	72	**161**	00:00:10:017	-242	-25	63
115	00:00:10:013	-237	-30	68	**162**	00:00:10:017	-243	-28	61
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124	00:00:10:014	-246	-32	45	**171**	00:00:10:018	-245	-27	58
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137	00:00:10:015	-251	-27	56	**184**	00:00:10:019	-247	-28	60
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142	00:00:10:016	-242	-28	58	**189**	00:00:10:020	-243	-25	61
143	00:00:10:016	-240	-26	58	**190**	00:00:10:020	-244	-28	61
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154	00:00:10:017	-244	-29	47
155	00:00:10:017	-243	-31	52
156	00:00:10:017	-243	-27	60
157	00:00:10:017	-242	-28	55
158	00:00:10:017	-246	-27	59
159	00:00:10:017	-240	-28	64
7	00:00:11:002	-252	-33	46
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9	00:00:11:002	-249	-24	50
10	00:00:11:005	-251	-17	49
11	00:00:11:005	-251	-17	56
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13	00:00:11:005	-240	-24	65
14	00:00:11:005	-241	-26	64
15	00:00:11:005	-236	-31	63
16	00:00:11:005	-231	-32	63
17	00:00:11:005	-237	-30	67
18	00:00:11:005	-242	-28	66
19	00:00:11:005	-246	-28	57
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21	00:00:11:005	-247	-32	49
22	00:00:11:005	-251	-29	45
23	00:00:11:005	-247	-32	42
24	00:00:11:005	-243	-36	43
25	00:00:11:005	-244	-31	42
26	00:00:11:006	-243	-35	47
27	00:00:11:006	-243	-27	42
28	00:00:11:006	-240	-31	61
29	00:00:11:006	-242	-32	64
30	00:00:11:006	-242	-27	62
31	00:00:11:006	-247	-28	73
32	00:00:11:006	-237	-18	63
33	00:00:11:006	-242	-29	65
34	00:00:11:006	-237	-33	63
35	00:00:11:006	-237	-31	58
36	00:00:11:007	-242	-38	56
37	00:00:11:007	-251	-28	54
38	00:00:11:007	-249	-31	41
39	00:00:11:007	-249	-27	48
40	00:00:11:007	-248	-27	50
41	00:00:11:007	-246	-27	53
42	00:00:11:008	-242	-25	54
43	00:00:11:008	-242	-26	58

198	00:00:10:021	-249	-29	56
199	00:00:10:021	-247	-33	59
0	00:00:11:001	-243	-29	55
1	00:00:11:002	-238	-27	57
2	00:00:11:002	-232	-21	64
3	00:00:11:002	-235	-24	56
4	00:00:11:002	-237	-32	55
5	00:00:11:002	-251	-37	49
6	00:00:11:002	-252	-42	46
54	00:00:11:008	-245	-29	54
55	00:00:11:008	-249	-27	56
56	00:00:11:008	-245	-27	57
57	00:00:11:008	-242	-28	61
58	00:00:11:009	-241	-27	56
59	00:00:11:009	-243	-28	59
60	00:00:11:009	-241	-30	62
61	00:00:11:009	-241	-30	63
62	00:00:11:009	-243	-27	63
63	00:00:11:009	-241	-30	63
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66	00:00:11:009	-244	-27	52
67	00:00:11:009	-244	-28	55
68	00:00:11:009	-244	-29	61
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70	00:00:11:009	-244	-28	53
71	00:00:11:009	-245	-27	58
72	00:00:11:009	-243	-25	51
73	00:00:11:009	-242	-30	56
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75	00:00:11:010	-242	-31	62
76	00:00:11:010	-243	-28	59
77	00:00:11:010	-242	-30	58
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80	00:00:11:010	-246	-28	57
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84	00:00:11:011	-243	-30	58
85	00:00:11:011	-242	-26	53
86	00:00:11:011	-246	-24	55
87	00:00:11:011	-244	-29	58
88	00:00:11:011	-242	-30	60
89	00:00:11:011	-245	-28	58
90	00:00:11:012	-244	-30	58

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46	00:00:11:008	-241	-29	63
47	00:00:11:008	-244	-29	64
48	00:00:11:008	-244	-29	56
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52	00:00:11:008	-248	-31	51
53	00:00:11:008	-247	-31	52
101	00:00:11:120	-235	-23	64
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103	00:00:11:120	-235	-27	59
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105	00:00:11:120	-249	-43	49
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107	00:00:11:121	-249	-34	44
108	00:00:11:121	-251	-24	47
109	00:00:11:121	-248	-23	54
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111	00:00:11:121	-246	-20	54
112	00:00:11:121	-244	-21	60
113	00:00:11:121	-239	-27	66
114	00:00:11:121	-239	-29	71
184	00:00:11:121	-235	-32	70
91	00:00:11:012	-242	-29	58
92	00:00:11:012	-242	-28	58
93	00:00:11:012	-243	-29	60
94	00:00:11:012	-242	-29	53
95	00:00:11:012	-243	-31	56
96	00:00:11:012	-248	-29	58
97	00:00:11:012	-250	-30	58
98	00:00:11:012	-250	-29	57
99	00:00:11:012	-246	-27	51
100	00:00:11:119	-238	-29	62

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