AN UWB RADAR BASED STEALTHY 'LIE DETECTOR'

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1. INTRODUCTION

The capability of remotely detecting breath rate and heart rate using radar techniques in the microwave region of the RF spectrum has been repeatedly demonstrated. Preliminary investigations[1, 2, 3], published as early as the 70's, described conventional short range radar methods capable of monitoring thorax movements and breathing activity as well. In the last few years, with the introduction of UWB radars, remote, non-contact vital parameters monitoring revived. Interesting medical applications of UWB radars were first described in the cardiologic-pulmonary fields[4, 5, 6] and in ENT (ear-nose-throat) clinics[7]. A more in-depth and wider presentation of the possible medical uses of UWB radars has recently been presented[8]. In principle, an UWB radar pulse is able to pass through the human thorax (no matter of clothing) so to be echoed back by the cardiac structures, namely the heart wall. The signal out of an UWB radar aimed at the subject's thorax is composed of two slow varying periodic signals: the first conveying heart movements and the other carrying information on thorax displacement and respiration rate. As such data carry pathological and physiological meanings to the physician, a new, UWB-based, vital parameters monitor has already been devised for a variety of applications in clinical, emergency and military medicine[8].

Heart and breath rates, along with blood pressure and electro-dermal activity, are also of importance in the so called "polygraph testing" more commonly known as lie detector test[9]. As a matter of fact "lie detectors" don't actually detect lies and there is also no widespread agreement if they might even detect a deceptive behavior of the examinee. Scientific evidence and admissibility in court of polygraph testing remain controversial. Many authoritative papers were presented, some of them giving a few credits for a limited scientific evidence[10] of the polygraph testing, others giving it no credits at all[11] even comparing it to astrology or tea-leaf reading[12]. Apart from the scientific validity of the

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polygraph testing *per se*, the added problem of countermeasures is actually a central one when the game becomes serious. Breathing and cardio countermeasures are in the training of every "professional" agent which might undergo polygraph testing under the enemy as well as under the friendly part. When properly mastered, countermeasures are much more "scientifically" effective than the polygraph itself[13]. To this end an improved "lie detector" should be invisible so that no countermeasures might be taken.

An UWB radar-based "lie detector machine" would add the incredible feature of "stealthiness" to the examination. As a matter of fact the subject under test might even be kept unaware of being polygraphed.

Present study was mainly directed to obtain a research instrument for fully legal psychophysiological studies in a clinical setting. The current research is aimed at developing devices to assess, in real time and with no active participation of the subject, the mental stress during every day activities like using a personal computer or driving. Final results should enable the design of new and ergonomically improved man-machine interfaces or improved safety devices in car driving (falling asleep alerts).

By the way, the use as a conventional "lie detector" is at present believed not allowed in the U.S., neither at federal nor at state level jurisdictions. As a matter of fact, both the U.S. Army Criminal Investigation Command and the Naval Investigative Service, for example, need a *Polygraph Examination Statement of Consent*[14] or a *Polygraph Examination Waiver*[15] signed by the prospective examinee. In those cases a stealthy polygraph would be actually unusable, not even for security screening purposes. Nevertheless a stealthy "lie detector" might be used in covert operations or, as in our case, for psycho-neurophysiological studies where the system is used on purpose without the subject's awareness. In this situation any subject's activity to counteract, fool or even be conditioned by the machine, should be avoided.

2. MATERIALS AND METHODS

The experimental setup is composed of an UWB radar device and an electrocardiograph (EKG) amplifier, for control purposes, connected to an analog to digital converter interfaced to a personal computer as depicted in figure 1.

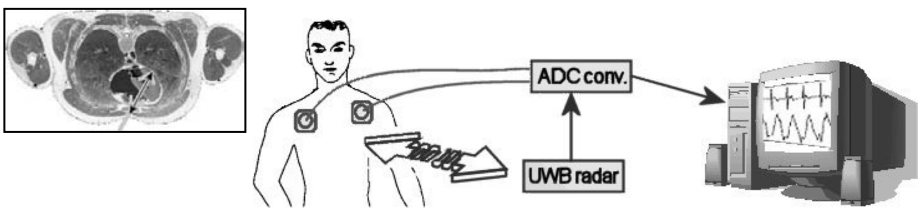
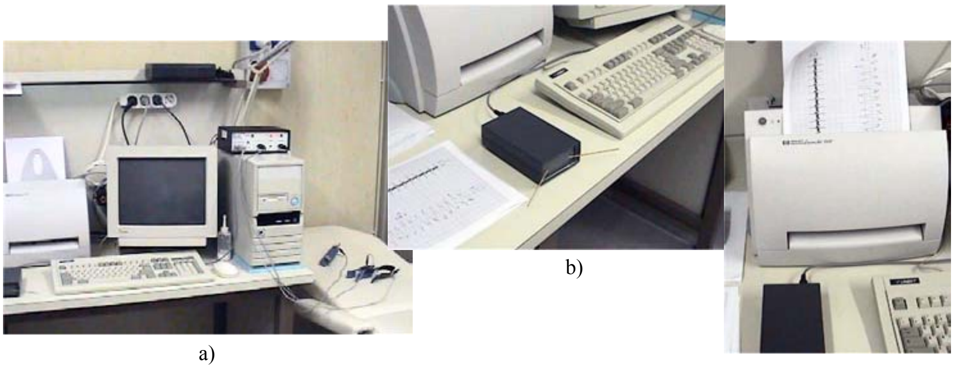


Figure 1. Schematic of the experimental setup for UWB vs. EKG comparison. Analog to digital converter (ADC conv.) includes a two channel EKG for detecting heart electrical activity at the same time with the UWB field disturbance signal. In the box on the left a slice from the Visible Human Project data[16] is shown with the possible radar beam path marked inside the thorax.

The analog to digital (ADC conv.) converter has been custom developed to integrate the UWB radar power supply, a dual channel electrocardiograph (EKG) and a multichannel, 10 bits resolution, 200 samples/sec/channel, analog to digital converter interfaced to the parallel port of a personal computer. The software was written in the Modula-2 programming language under the MS-DOS operating system. It dates back to the 90's and it has been used in our laboratory since then as a standard for research purposes on dedicated computers.



a) b) c)

Figure 2. Experimental setup from left to right: a) hardware configuration on the bench; analog to digital converter and EKG amplifier on top of the personal computer; b) UWB radar prototype with dipole ribs protruding out of the black box; c) printer.

The UWB radar device used in this work is a replica with modifications of the micropower-impulse radar (MIR) originally developed at Lawrence Livermore National Laboratory (LLNL) in 1993[5, 6, 17]. Actual modifications and a detailed practical implementation of a prototype of the device has been presented elsewhere[18]. The UWB radar has been prototyped on a multi-hole board and emits 2 ns time-width pulses through a simple dipole antenna with considerable ringout. Although the assembly was all but adequate for a microwave system, nevertheless it was very successful in the preliminary investigations for a possible medical use. Pulses are emitted with a randomly variable pulse repetition delay (random dithering) having a mean repetition frequency of 2 MHz. Actual duty cycle is about 0.4% (not considering the ringing).

3. PRINCIPLE OF OPERATION

Figures 3 and 4 describe UWB radar signal generation from a moving reflector like the heart wall. In practice an analog signal is obtained thanks to the rise-time of the electromagnetic pulse. The rising shape of the reflected pulse is sampled at different time instants according to the time-delay in the pulse round trip. Although it is more commonly known as a "radar", from the above rough presentation of its working principle, the kind of system presented here should be more appropriately named a "field disturbance sensor" or, generically, an "electromagnetic sensor".

In this regard the UWB radar is definitely not a brand new biomedical microwave sensor in respect to microwave CW devices described in the past, which used interferometry[19], Doppler effect[20, 21, 22], backscattering[23, 24], or plain reflection[25]. As a matter of fact the definition of "field disturbance sensor" has been used in the latest Federal Communications Commission (FCC) Report and Order on the revision of Part 15 of the Commission's Rules regarding Ultra-Wideband Transmission Systems[26].

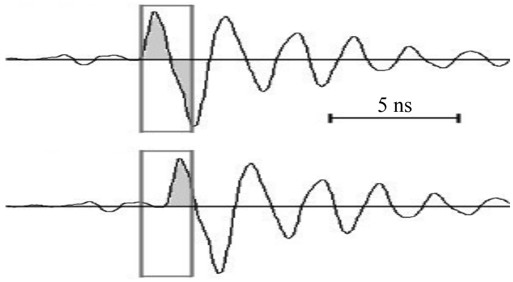


Figure 3. Example of UWB radar signal sampling for an UWB radar in monostatic mode $5 \cdot 10^{-2}$ m (2") away from the anterior thorax wall. Upper trace: echo from the heart wall in diastole, lower trace: echo from the heart wall in systole. Echo is delayed in systole as the heart wall is globally more distant (inside the thorax) from the antenna. Aperture time of the receiving sampler, in the boxes, is shown at the same time delay after Tx pulse emission). Echo signal in the aperture time (averaged over many pulses) constitutes the signal output of the UWB radar.

In light of the cited FCC document the present device is all but legal as it doesn't comply with the rules. In effect, although no precise determination of frequency band of emission and no clear determination of spurious emissions has been carried out yet, nevertheless the device should be working in the low frequency band (below 960 MHz, as per section 15.509 of FCC rules) were medical imaging devices are not allowed.

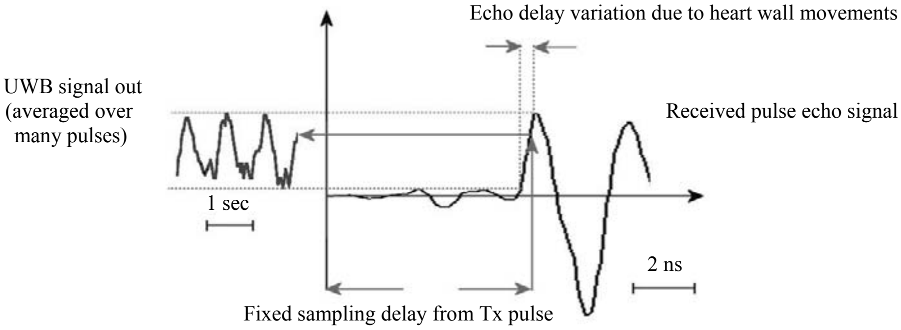


Figure 4. Genesis of the UWB heart signal exemplified considering an instantaneous sampling aperture. Although pulses are emitted with random time delay repetition, sampling is always synchronous with the transmitted pulse. UWB signal comes out of noise after thousands of averaged samplings (note the different time scales). Random dithering and subsequent averaging cancel heterodyning with other continuous wave (CW) signals present on the antenna.

According to the FCC, UWB based medical devices are only allowed in the high frequency band from 3,100 MHz to 10,600 MHz (as per section 15.513 of newly adopted FCC rules). However, the problem of compliance, with FCC or ERO rules and with any prior intellectual property issue, will be addressed at the time, if any, where the present research prototype will be engineered for industrial or commercial purposes.

A few more considerations would be useful for a better understanding of the operation and design trade-offs.

From the signal generation scheme presented in figure 4 it should have to be noted that a conversion is made of heart wall displacement to output voltage thanks to echo pulse rise time. In fact, for a fixed sampling time delay from Tx pulse emission, target surface displacement, variable in time, is responsible for a phase shift in the echo so that sampling occurs at different levels in the pulse rising period. This effect implies that there exists a minimum rise time in the pulse for completely acquiring the full dynamic range of the target moving surface (the heart wall). For example, estimating an electromagnetic field propagation speed in the tissues at about $180 \cdot 10^{6}$ m/s (some $400 \cdot 10^{6}$ miles per hour) and a maximum displacement of the heart wall during heart contractions in the range of $15 \cdot 10^{-3}$ m (half an inch), a minimum rise time, for the pulse, results at about $90 \cdot 10^{-12}$ sec or 90 ps. This extreme case is exemplified in figure 4 for didactic purposes. A shorter rise time would have the effect of sensing a given surface more than once and the "onion effect" will be shown due to the presence of many pulses (ringing) in space at the same time. This might give an erroneous output signal presenting multiple waves per cardiac cycle as already hypothesized in a previous work[5]. It should effectively lead to the determination of a heart beat frequency erroneously multiple of the real one[5].

It is often declared, when talking about UWB radio technology, that this technique is highly immune to multi-path fading or to its radar counterpart, the clutter. This is absolutely not the case with ringing pulses. As the emitted pulse is larger than irradiated structures, not only diffraction plays a major role, but summation of echoes at the receiving antenna is of the highest importance. If the echoes which sum up at the input are coming back from still targets and only one is coming from the moving heart wall, than a simple and inoffensive, although quite high, d.c. level should be expected over the useful signal. In practice, echoes arriving from moving structures like lungs, blood vessels and thorax wall produce the effect of summing their signals (with various and time varying amplitudes) to the expected echo signal bouncing back from the target heart wall surface. No matter if the reflecting surfaces are before or behind the target. Furthermore, the echoes arrive at the antenna (in a true monostatic mode with a single Tx-Rx radiator) when the emitted ringing pulse is not yet completely gone away. So the sampler is compelled to acquire part of the Tx pulse itself and this adds another exceptionally high d.c. level to the output signal. To cope with this effect an high pass filter has to be added in the amplifier chain as early as possible after the fast echo sampler. This feature only enables the reaching of the high gain needed to rise the signal to acceptable levels, without saturating the amplifier. From the previous consideration, it should emerge the rationale of naming this kind of UWB radar as a "field disturbance sensor", more appropriately.

The sampling of its own emitted pulse, yet on the antenna when useful echo arrives, makes also impossible to perform a fast scan of targets in space by varying the sampling delay. In fact, as a long time constant for the high pass filter will be chosen, so to have an high pass at low frequency obtaining a good tracking of low speed of displacement targets, the scanning in space (that is the continuous variation of sampling delay from Tx pulse emission) should have to be set even slower. In this case only, indeed, the resulting spurious d.c. level variation (caused by the sampling of the emitted ringing pulse at different places) will be effectively canceled out by the high pass filter. In the experiments performed so far not a single adequate trade-off was found in the latter regard, and scanning in space eventually considered impossible with present device. This effect might limit seriously any possibility to consider an UWB radar device for a real world medical (ultrasound-like) imaging scanner.

On the other end, a low pass filter have to be used too, for cleaning the UWB radar signal from considerable noise. This limits the tracking of high speed of displacement targets as the heart valves are. By the way, technology improvements and trade-offs are possible in this respect.

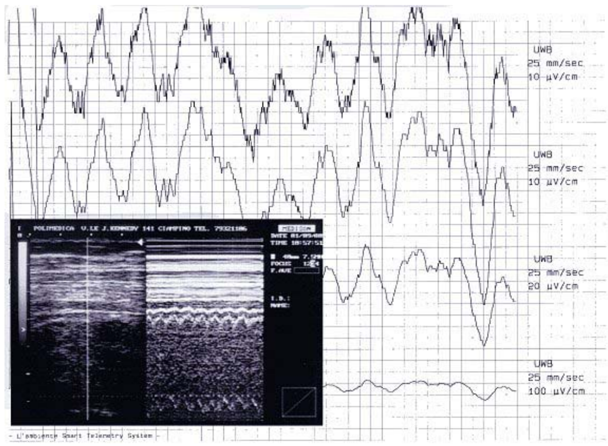


Figure 5. In the tracings the same UWB heart echo signal, with various degrees of low pass filtering and amplification, is plotted. Breathing modulation is visible due to movements of the chest wall. In the insert (bottom-left) an actual ultrasound echo image of the chest in the same subject is shown (SONOACE 1500 scanner from MEDISON Co.). On the right the M-mode tracing shows the heart wall displacement, approximately in the center of the image. Thorax movements due to breathing are not visible as the ultrasonic transducer, obviously, moves along with the chest. A good match is demonstrated between ultrasound M-mode echo and UWB M-mode echo.

As regards signal loss in the pulse round trip, it should be expected, on a pure time domain reflectometry theory[27], that the return loss be due both to impedance mismatch and to attenuation into lossy media. Measurements and models, based on actual electromagnetic data of human tissues[28], gave return loss figures, for a two way monostatic radar, in the order of 60 dB. Comparable results were derived from previous studies on UWB propagation through lossy media[29]. It appears that, for very short range radar operation, the R^{-2} dependence of power density on target distance R can be omitted and conventional two way radar equation, commonly found on reference handbooks[30], seems at present not adequate to model radar propagation in this very special situation.

4. PHYSIOLOGICAL CONSIDERATIONS

From the previous description it appears that an UWB radar based monitor is able to detect heart and breath rate only. In conventional "lie detectors" attention is also given to monitor other signals capable of giving an insight of autonomic nervous system modifications which might be used to infer a deceptive behavior of the subject.

Parasympathetic and sympathetic sections of the autonomic nervous system in humans play an important role in the control of heart rate and, from proper elaboration of the heart rate variability (HRV) signal, interesting data can be obtained.

HRV is obtained by measuring time interval between successive heart beats. Using EKG to detect heart beats, HRV is also known as "R-R signal" as time intervals are actually measured between successive R-waves in the EKG. Using UWB radar to detect heart beats, a suitable hardware or software detector measures time intervals between successive peaks or valleys in the radar signal (head to head or foot to foot in the UWB radar wave like that depicted in figure 5).

HRV is fundamentally different from a signal derived by sampling a continuous process. HRV is by default a sampled signal whose sampling period is the signal itself. Although there is no sound physiological reason for considering HRV coming from a continuous process (i.e. like the output of a delta-modulator), nevertheless it is common practice to derive a uniformly sampled HRV, from crude HRV, by interpolation (as if HRV might be defined for all points in time). In this way standard analyzing techniques can be used. HRV has been studied in the last twenty years in heart research labs worldwide and it is now used, more and more frequently, on a clinical basis.

Continuous alteration in sympathetic-parasympathetic balance of the autonomic neural regulation makes the heart rhythm fluctuate around the mean heart rate. These fluctuations are periodic or quasi-periodic and they seem to originate from respiration regulation, blood pressure and thermoregulation. Parasympathetic (vagal) regulation of the heart is inhibited during inspiration[31] and it can be excluded using antivagal drugs like atropine[32]. Oscillations in blood pressure, due to baroreflex control loop, result in sympathetic activation[33]. Changes in peripheral blood vessel resistance due to thermoregulatory adjustments of peripheral blood flow are under the control of sympathetic activation too[33]. Parasympathetic stimuli make the heart to slow down while sympathetic activation makes the heart rate to increase.

Under the hypothesis that mental stress could interact with sympatho-vagal system, commonly used "lie detectors" monitor the electrodermal activity. In our case HRV is monitored instead. Mental stress (for example in mental arithmetic) has been found to increase sympathetic activity and decrease parasympathetic activity as detected using HRV analysis[34]. Studies were also conducted to demonstrate the usefulness of HRV analysis to detect negative functional states (asthenia, depression, neurotic symptoms) in the course of psychological relaxation sessions[35].

Keeping in mind the fact that no true and scientific "lie detector" exists or can be developed within the present state of the art, we can nevertheless consider HRV, detected with an UWB based radar, as a viable way of obtaining a signal related to mental stress. This might prove useful in situations where the autonomic nervous system of the subject is involved.

5. RESULTS

In figure 6 a typical plotting of simultaneously recorded EKG and heart UWB radar signals is presented. The UWB radar signal, related to mechanical movement of the heart, follows the electrical event by a small amount of time as expected.

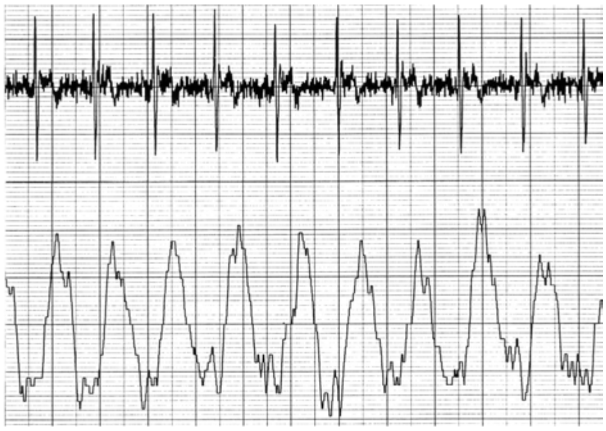


Figure 6. UWB radar tracing (bottom trace 20 μV/cm) versus high-pass filtered ECG (upper trace 400 μV/cm), time base 15 mm/sec.

UWB radar signal of heart displacement is quite noisy although very clearly and easily detected from a distance of $5 \cdot 10^{-2}$ m (2") from the thorax. In the initial experiments a thick plastic sheet was fixed to the table and the volunteer subject asked to press his chest onto the plastic sheet. In this way skin movements were strongly avoided and sufficient confidence was obtained that the UWB echoes are coming from inside the body. Subsequent signals were obtained with the antenna placed on the rear part of the thorax and the subject asked to seat comfortably on a chair. UWB radar worked at a distance of $15 \cdot 10^{-2}$ to $20 \cdot 10^{-2}$ m from the heart, with the back of the chair in between, as depicted in figure 7.

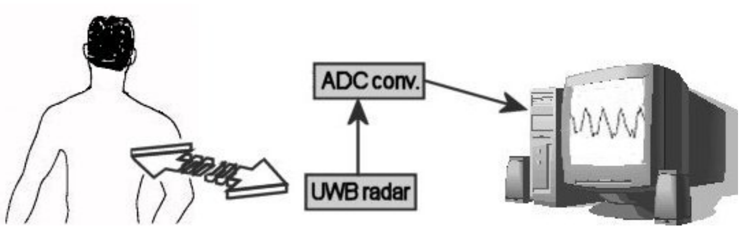


Figure 7. Schematic of the stealthy vital parameters monitor to be mounted into the back of a chair.

On the tracings in figure 6, two different heart cycle detectors implemented in software were used to deliver heart beat period from EKG signal and UWB echo signal as well. The EKG signal was high pass filtered and R wave was then detected using dynamic thresholding. Threshold is updated to the EKG signal whenever the latter is higher than the threshold. During time intervals in which EKG signal is lower than the threshold, then its level is decreased in an exponential way so to cope with varying R wave amplitudes in successive beats. This method proved quite robust although somewhat *naïve*. Heart beat period detection on UWB echo tracings used an algorithm previously used for arterial pressure analysis. Peaks and valleys of the signal are detected and time lag between two consecutive valleys is considered the heart beat period. This kind of processing is quite sensitive to noise and it will be certainly abandoned in further studies. It was used at this moment just because it was ready, having been developed for other purposes.

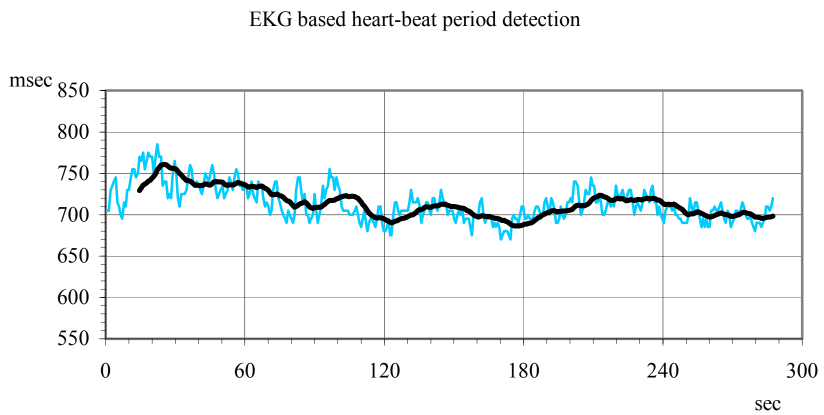


Figure 8. Heart cycle period estimated from EKG tracing during five minutes (400 heart beats). Superimposed a 20 points box-car moving average regression line is plotted.

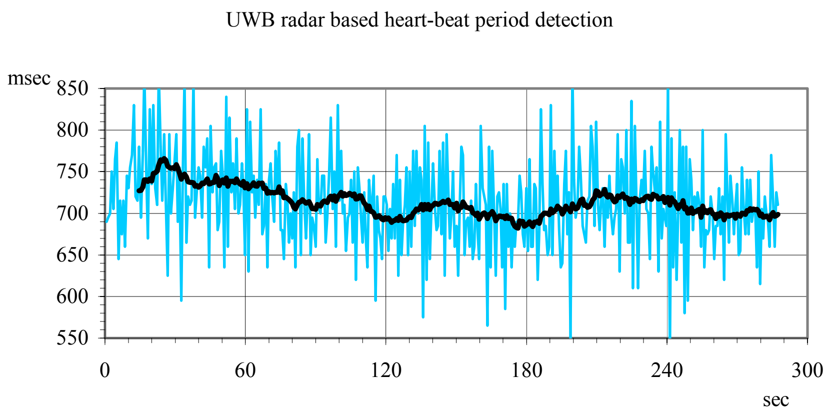


Figure 9. Heart cycle period estimated from UWB radar echo tracing during five minutes (400 heart beats). Superimposed a 20 points box-car moving average regression line is plotted. Data in this figure and in the preceding figure were recorded simultaneously on a subject at rest.

A simple statistical analysis of the data gave a mean heart period for EKG detected beats of 713.1 ms and a mean heart period for UWB detected beats of 712.9 ms. As both detection algorithms detected without errors each heart beat (no heart beat was missed), the same mean value was expected. Standard deviation differed significantly in the two series of data. In EKG derived data standard deviation was 21.06 while in UWB derived data it was 56.95. The difference in standard deviation is to be accounted for the higher noise which contaminated UWB derived HRV signal. Histogram comparison in figure 10 demonstrates this phenomenon.

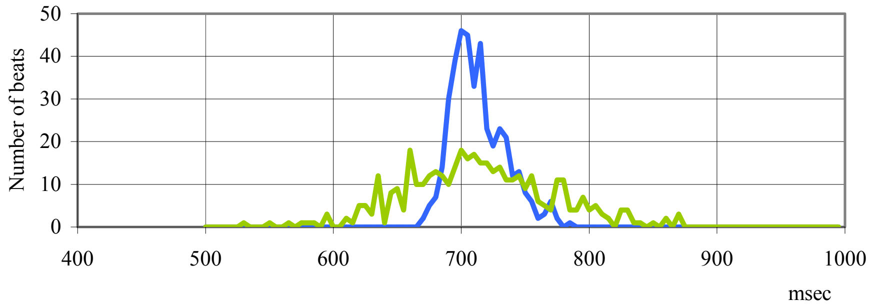


Figure 10. Histograms of heart periods detected using EKG (darker line) and UWB (lighter line). The histograms plot the number of heart beats having the same period (no matter they were consecutive or not). Noise in the UWB echo tracing causes errors in period estimation and spreading of the relative histogram while both distributions peak at the same mean value.

As explained in the previous paragraph, autonomous nervous system activation should determine cyclic modifications on the heart period. Spontaneous respiration rate should account for variations intervening at breath rate, sympathetic blood pressure variations and thermoregulation should also account for cyclic behavior at lower frequencies.

To verify previous finding reported in the literature, EKG derived HRV signal was Blackman-Harris windowed[36] and a spectra was estimated using Fast Fourier Transform (FFT).

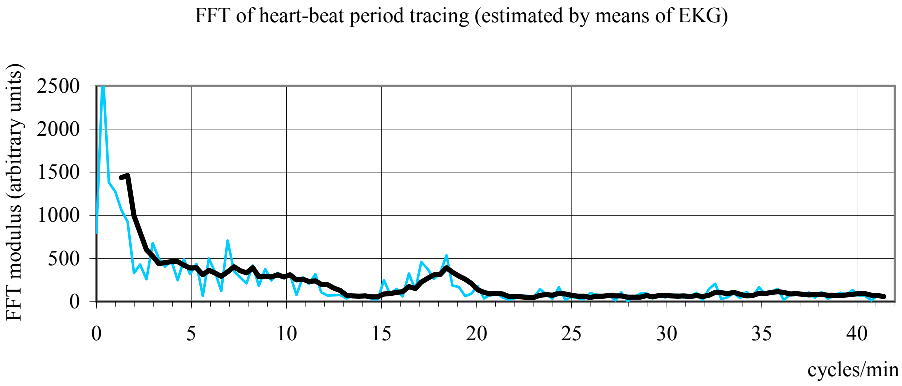


Figure 11. Spectrum of HRV signal derived from EKG (R-R tracing). Box car 5 points moving average regression line is superimposed for a clearer understanding.

A mean "sampling rate" was considered which is equal to the inverse of mean heart beat period, i.e. 1/0.713 Hz or 1.4 Hz. A 256 point FFT yielded a spectrum with a resolution of 5.4 mHz or 0.33 cycles/min approximately. The same was performed on UWB echo derived HRV signal. In figure 11 the spectrum of the HRV signal is plotted and a box-car moving average regression line is superimposed on. Breath rate related variation is clearly visible in the 15-20 cycles/min band. Sympathetic activation is also visible in the 5-12 cycles/min frequency band. It is important to note that such periodic behavior of HRV signal was also appreciable by naked eye in the tracings reported in figure 8 and 9 from which spectra were estimated. The highly noisy UWB radar based HRV signal is absolutely not suitable for FFT processing as Fourier Transform looses meaning with uncorrelated noise build-up in the original signal as visible in figure 12.

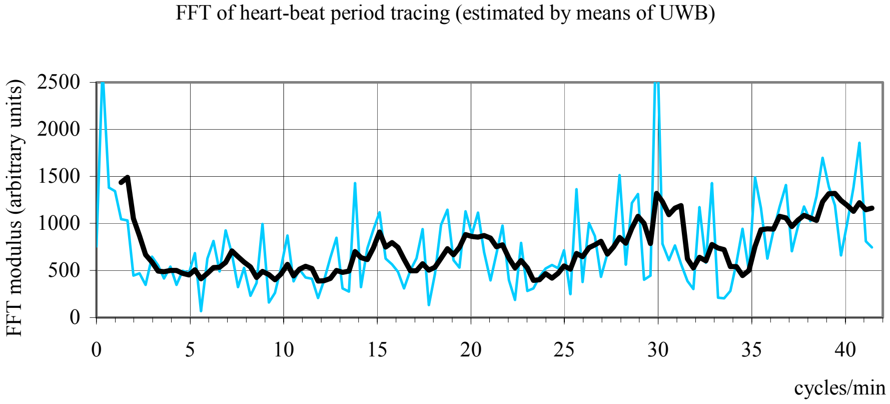


Figure 12. Spectrum of HRV signal derived from UWB radar echo tracing. Box car 5 points moving average regression line is superimposed for a clearer understanding. With heavy noise contamination, Fourier Transform becomes statistically unsignificant. The spectrum is meaningless in this case.

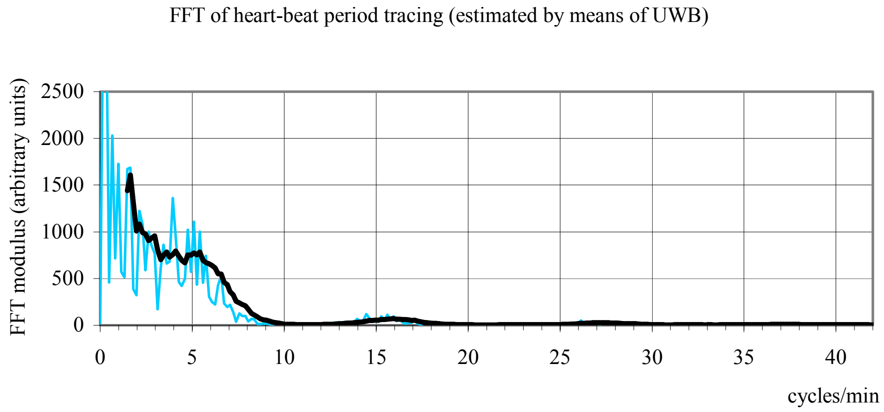


Figure 13. Spectrum of HRV signal derived from UWB radar echo tracing after Wavelet denoising. After denoising breath rate related peak is no more visible. Sympathetic activation is very low in the 3-10 cycles/min frequency band.

To cope with this problem HRV frequency estimation is more commonly performed using autoregressive modeling (AR) or autoregressive moving average (ARMA) methods. Non linear methods are being used too.

By the way, it is curious how distant bands in the spectrum are used in this work: from GHz in the microwave region of UWB radio to mHz in the heart rate variation region.

In this initial study decision was taken to filter out noise in the UWB based HRV signal using Wavelet based filtering[37] and then use FFT on the recovered signal. This procedure is somehow arbitrary, nevertheless it was quite effective in cleaning out noise preserving low frequency spectral information. Discrete 256 points Wavelet transform was calculated and decimation of higher resolution scales in the Wavelet domain was carried on. After inverse Wavelet transform a conventional 256 point FFT was used to estimate the spectrum. A more strict procedure is planned for future research.

To test the system, a math game was prepared and administered to a voluntary subject. Randomly generated arithmetic expressions were presented on the screen of the computer and the subject was asked to mentally do the calculation and enter the correct result as soon as possible.

The situation was made quite challenging as the calculations were programmed more difficult as the time passing by. After an initial period in which the subject feels he is adequate to the task and he is relaxing, an increase in difficulty of proposed calculations introduces anxiety and fear in the subject. Sympathetic activation arises and it is clearly visible in figure 14 as respect to figure 13.

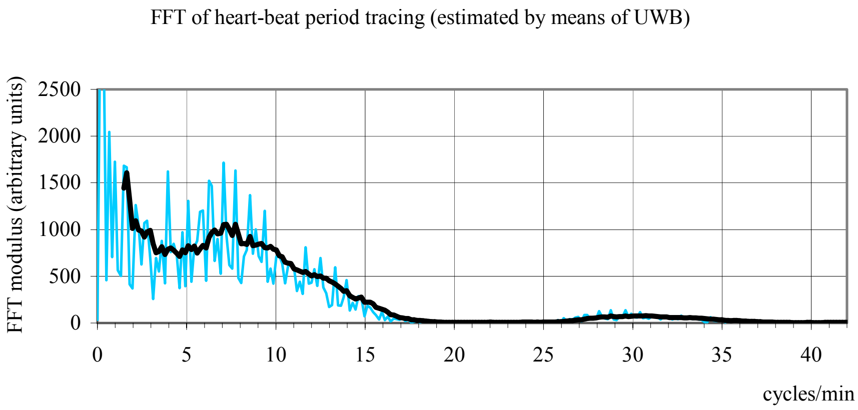


Figure 14. Spectrum of HRV signal derived from UWB radar echo tracing after Wavelet denoising with breath rate related peak no more visible. A clear peak in the 5-10 cycles/min band is visible due to sympathetic activation caused by a math game (doing mental math calculations as soon as possible).

This findings were already reported in the literature and the system testing was started in this way looking for confirmation of previous results. In fact, main objective of the task was to prove the equivalence of UWB radar derived HRV signal with the EKG derived one.

A straightforward application of this technique, thanks to its intrinsic remoteness and exceptional userfriendliness, is the monitoring of car drivers to prevent they falling asleep during driving. A small UWB radar attached on, or built into, the driver's seat safety belt could sense heart rate and the HRV signal to deliver a suitable alarm when necessary.

Figure 15. An UWB radar based heart rate monitor mounted into the driver's safety belt to prevent the driver falling asleep.

The issue of safety, when using electromagnetic energy, is often of primary concern. Studies on the effects of UWB electromagnetic energy on humans, animal models and living cell cultures were performed in the past[38, 39, 40, 41, 42, 43]. Unfortunately their scope was primarily addressed towards the assessment of the possible side effects on humans of UWB electromagnetic weapons. To this end power in excess of megawatt was used along with UWB electric fields in the range of tens to hundreds of kV/m. Although no final conclusion can be obtained from the limited published reports, nevertheless in our case power level and electric field intensity are seven to nine orders of magnitude lesser and no data at all is available from the literature in this regard.

Nowadays, the only argument that can be given in favor of the safety of UWB radar based heart rate monitors is that whatever dangerous cellular phones might be, medical UWB radars emit two to three orders of magnitude less power than common cellular phones. This means that a five minutes phone call on a mobile might equal a ten hours UWB radar heart rate monitoring.

6. CONCLUSIONS

The viability of monitoring sympathetic/parasympathetic activation due to mental stress in humans using the heart rate variability signal (HRV) detected with an UWB radar based vital signs monitor has been assessed. The system's strength is based on the possibility to maintain the subject under test unaware of being monitored so to avoid any psychological discomfort. This leads to a more objective measure which, in the case of a proposed new "lie detector machine", adds the extra bonus of avoiding countermeasures when operated in a stealthy mode. The main weakness of the system lies in the presence of noise which strongly degrades beat to beat time period detection using an automatic software heart cycle detector. Although advanced filtering techniques effectively help in overcoming noise disturbance, nevertheless a better software detector might solve the problem in a more robust way.

Apart from "lie detectors", the proposed system and signal processing software may be more appropriately used in neuro-psychological studies aimed at developing better man-machine interfaces or in detecting the degradation of human performances in dangerous or risky tasks like car driving, aircraft piloting or critical monitoring tasks.

7. ACKNOWLEDGEMENTS

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8. REFERENCES

1. C. G. Caro, and J. A. Bloice, Contactless apnoea detector based on radar, *Lancet* 2(7731), 959-961 (1971).
2. C. I. Franks, B. H. Brown, and D. M. Johnston, Contactless respiration monitoring of infants, *Med. Biol. Eng* 14(3), 306-318 (1976).
3. C. I. Franks, J. B. Watson, B. H. Brown, and E. F. Foster, Respiratory patterns and risk of sudden unexpected death in infancy, *Arch. Dis. Child* 55(8), 595-604 (1980).
4. S. G. Azevedo, and T. E. McEwan, Micropower Impulse Radar, *Energy and Technology Review*, UCRL-52000-96-1, 1-2 ,16-13 (1996).
5. T. E. McEwan, Body monitoring and imaging apparatus and method, United States Patent 5,573,012, (1996).
6. T. E. McEwan, Body monitoring and imaging apparatus and method, United States Patent 5,766,208, (1998).
7. J. F. Holzrichter, G. C. Burnett, L. C. Ng, and W. A. Lea, Speech articulator measurements using low power EM-wave sensors [letter], *J. Acoust. Soc. Am.* 103(1), 622-627 (1998).
8. E. M. Staderini, UWB Radars in Medicine, *IEEE AESS Systems Magazine* Jan 2002, 13-18 (2002).
9. K. Bonsor, How lie detectors work, http://www.howstuffworks.com/lie-detector.htm/printable (as visited on Jan 2002).
10. *Scientific Validity of Polygraph Testing: A Research Review and Evaluation*, Technical memorandum, Washington D. C.: U.S. Congress Office of Technology Assessment OTA-TM-H-15 (1983).
11. G. W. Maschke, and G. J. Scalabrini, *The Lie Behind the Lie Detector,* http://antipolygraph.org (as downloaded on Jan 2002).
12. D. C. Richardson, Opening Statement before the U.S. Senate Committee on the Judiciary's Subcommittee on Administrative Oversight and the Courts, http://www.nopolygraph.com/drewtest.htm (as visited on Jan 2002).
13. D. G. Williams, *How to Sting the Polygraph*, Chickasha, Oklahoma: Sting Publications (1996).
14. U.S. Army criminal investigation command, Polygraph examination statement of consent, http://www.fas.org/sgp/othergov/polygraph/ota/consent2.jpg (as downloaded on Jan 2002).
15. Naval Investigative Service, Polygraph examination waiver, http://www.fas.org/sgp/othergov/polygraph/ota/consent1.jpg (as downloaded on Jan 2002).
16. Chang Y.J., *The NPAC Visible Human Viewer*, Syracuse University, NY. http://www.npac.syr.edu/projects/vishuman/VisibleHuman.html (as downloaded on Jan 2002).
17. T.E. McEwan, Ultra-wideband radar motion sensor, United States Patent 5,361,070, (1994).
18. E.M. Staderini, Everything you always wanted to know about UWB radars…, Online Symposium for Electronics Engineers. Web broadcasted on Jan 16th, 2001. http://www.osee.net/whitepapers/paper4.pdf (as visited on Jan 2001).
19. D.W. Griffin, MW interferometers for biological studies, *Microwave Journal*, May 1978, 69-72 (1978).
20. K.H. Chan and J.C. Lin, An algorithm for extracting cardiopulmonary rates from chest movements, *IEEE/Seventh Annual Conference of the Engineering in Medicine and Biology Society*, 466-469 (1985).

21. J.Y. Lee and J.C. Lin, A microprocessor-based noninvasive arterial pulse-wave analyzer, *IEEE Transactions on Biomedical Engineering*, BME-32, 6, 451-455 (1985).
22. W. Byrne, P. Flynn, R. Zapp and M. Siegel, Adaptive filter processing in microwave remote heart monitors, *IEEE Transactions on Biomedical Engineering*, BME-33, 7, 717-722 (1986).
23. K. Chen, D. Misra, H. Wang, H. Chuang and E. Postow, An X-band microwave life-detection system, *IEEE Transactions on Biomedical Engineering*, BME-33, 7, 697-701 (1986).
24. J.C. Lin, Microwave sensing of physiological movement and volume change: a review, Bioelectromagnetics, 13, 557-565 (1992).
25. K. Chen, Y. Huang, J. Zhang and A. Normal, Microwave life-detection systems for searching human subjects under earthquake rubble or behind barrier, *IEEE Transactions on Biomedical Engineering*, 27, 105-114 (2000).
26. *First Report and Order in the matter of Revision of Part 15 of the Commission's Rules Regarding Ultra-Wideband Transmission Systems*, ET Docket 98-153, Federal Communication Commission Washington, D.C. 20554; released April 22, 2002.
27. Hewlett Packard, *Time domain reflectometry theory*, Application Note 1304-2.
28. C. Gabriel, *Compilation of the Dielectric Properties of Body Tissues at RF and Microwave Frequencies*, Physics Department, King's College London, London WC2R 2LS, UK. Armstrong Laboratory (AFMC), Occupational and Environmental Health Directorate, Radiofrequency Radiation Division, 2503 D Drive, Brooks Air Force Base, TX, 78235-5102. Report: AL/OE-TR-1996-0037 (1996).
29. M.A. Barnes, L.W. Fullerton and P. Withington, *Pulson lossy media communication link project*, Pulson Communication Inc., Atlanta (GA) 30339 (report to the FCC).
30. *Electronic warfare and radar systems engineering handbook*, NAWCWPNS TP 8347, Naval Air Warfare Center Weapons Division Avionics Department Electronic warfare Division Point Mugu, CA 93042 (1999).
31. D. Eckberg, Human sinus arrhythmia as an index of vagal cardial outflow, *Respirat. Environ. Exercise Physiol.* 54, 4, 961-966 (1983).
32. S. Akselrod, D. Gordon, J. Madved, N. Snidman, D. Shannon and R. Cohen, Hemodynamic regulation: investigation by spectral analysis, *Am. J. Physiol.* 249, 867-875 (1985)
33. C. VanRavenswaaij-Arts, L. Kollée, J. Hopman, G. Stoelinga and H. VanGeijn, Heart rate variability, review, *Ann. Intern. Med.* 118, 6, 436-447 (1993).
34. M. Pagani, R. Furlan, P. Pizzinelli, W. Crivellaro, S. Cerutti, and A. Malliani, Spectral analysis of R-R and arterial pressure variabilities to assess sympathovagal interaction during mental stress in humans, *J. Hypertens. Suppl.* 7(6), S14-5 (1989).
35. Mashin V.A. and Mashina M.N., Analysis of the heart rate variability in negative functional states in the course of psychological relaxation sessions, Human Physiology, 26, 4, 420-425 (2000). Translated from Russian originally published in Fiziologiya Cheloveka, 26, 4, 48-54 (2000).
36. F.J. Harris, On the Use of Windows for Harmonic Analysis with the Discrete Fourier Transform, *Proceedings of IEEE*, January 1978.
37. M. Akay (ed.), *Time Frequency and Wavelets in Biomedical Signal Processing*, IEEE Press Series on Biomedical Engineering, John Wiley & Sons, Inc. (1998).
38. S.F. Cleary, F. Nickless, L.M. Liu and R. Hoffman, Studies of exposure of rabbits to electromagnetic pulsed fields. *Bioelectromagnetics*, 1, 3, 345-352 (1980).
39. J.A. D'Andrea, B.L. Cobb and J.O. Lorge, Lack of behavioral effects in the rhesus monkey: high peak microwave pulses at 1.3 GHz, *Bioelectromagnetics*, 10, 1, 65-76 (1989).
40. T.J. Walters, P.A. Mason, C.J. Sherry, C. Steffen and J.H. Merritt, No detectable bioeffects following acute exposure to high peak power ultra-wide band electromagnetic radiation in rats, *Aviat. Space Environ. Med.*, 66, 6, 562-569 (1995).
41. C.J. Sherry, D.W. Blick, T.J. Walters, G.C. Brown and M.R. Murphy, Lack of behavioral effects in non-human primates after exposure to ultrawideband electromagnetic radiation in the microwave frequency range, *Radiat. Res.*, 143, 1, 93-100 (1995).
42. J.H. Merritt, J.L. Kiel and W.D. Hurt, Considerations for human exposure standards for fast-rise-time high-peak-power electromagnetic pulses, *Aviat. Space Environ. Med.*, 66, 6, 586-595 (1995).
43. J.R. Jauchem, R.L. Seaman, H.M. Lehnert, S.P. Mathur, K.L. Ryan, M.R. Frei and W.D. Hurt, Ultra-wideband electromagnetic pulses: lack of effects on heart rate and blood pressure during two-minute exposures of rats, *Bioelectromagnetics*, 19, 5, 330-333 (1998).