SENSOR INTEGRATION IN AN RFID TAG FOR MONITORING BIOMEDICAL SIGNALS

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Abstract

RFID is evolving as a major technology enabler for identifying and tracking goods. RFID applications in biomedical area not only need to detect but also require monitoring and transmitting vital signals like heartbeat. The basic building blocks of an RFID tag for biomedical application are studied. This paper mainly focuses on the design and optimization of sensor that could be easily integrated in an RFID tag for remote monitoring of heartbeat signal. The sensor is based on the detection of capacitance variation when a heartbeat signal pressure is applied. The characteristics of the sensor are presented and this work would be very relevant for the sensing, remote monitoring of biomedical signals.

Keywords: RFID Tag, Biomedical application, Capacitive pressure sensor.

1. Introduction

Radio Frequency Identification (RFID) is based on remotely retrieving information via radio waves from miniature electronic circuit tags. Although the principle was established in late 40's it is only recently that the technology has taken off due to decrease in cost and increased capability [1]. RFID is evolving as a major technology enabler for identifying and tracking goods and assets around the world. It can help hospitals to locate expensive equipment more quickly, to improve patient care, to reduce counterfeiting and logistics providers to improve the management of moveable assets [2]. It also promises to enable new efficiencies in the supply chain by tracking goods from the point of manufacture through to the retail point of sale [3].

All these above-mentioned applications are mainly for identification purpose, but RFID in biomedical application requires not only identification but also monitoring of some vital signals in living creatures. Implanting a tag inside the living creature will enable us to monitor vital signals in addition to identification. An implanted sensor allows one to obtain data from inside the body without having to deal with wires or tubes, which penetrates the dermal

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(skin) barrier. This reduces the possibility of infection, which is frequently a greater contributor to morbidity and/or mortality [4]. Some of the biological signals like heartbeat, ECG etc. could be measured by placing the tag externally also [5]. Heartbeat detection and transmission is taken as specific example in this work.

The paper mainly focuses on the design and optimization of sensor integrated in an RFID tag for remote monitoring of biomedical, heart beat signal. The paper is organized in the following way; the basic building blocks of an RFID for biomedical application are covered in the immediate section. The sensor design and optimization is presented in section 3. The results are discussed in section 3.3 and finally the sensor is integrated with the remaining blocks and concluded.

2. Building blocks of RFID tag for biomedical application

The basic block diagram of an RFID tag for a general biomedical application is given in Fig. 1. The functions of individual blocks are presented in the following subsections

2.1. Sensor block

The blood flow through the veins can be detected by a sensor placed in a convenient part of the body, for e.g. on wrist. The sensor block converts the blood flow into heartbeat signal, which is the electrical equivalent of the physiological signal. This conversion can be achieved with the help of the designed pressure sensor described in section 3.

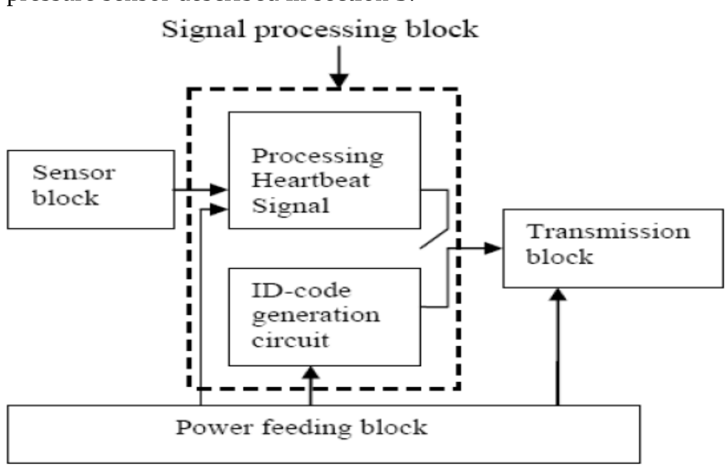

Fig. 1. Basic Building Blocks of an RFID tag for Biomedical Application

2.2. Signal processing block

The signal processing block consists of the heartbeat signal processing unit and the ID-code generation unit. The signals sensed by the sensors are weak and they need to be amplified. The signals also contain broadband noise component due to the other body activities and these noises can be eliminated using suitable circuits [6]. The signal processing block also includes circuits to digitize the heartbeat signal. The resulting signal gives the heart rate of the living being in which the tag is implanted. Each Tag has its own ID-code and the code is transmitted for the identification purpose. The code is stored in a ROM whose size depends on the size of ID-code. The code and the digitized heartbeat signals are sent to the transmitting block via appropriate digital circuits [7][8].

2.3. Transmitting block

The processed signal is to be transmitted to the reader. For low-voltage and low-power realization of a digital modulator, BASK (Binary Amplitude Shift Keying) modulation can be adopted. To realize BASK modulation, ON/OFF switching of a loop of a ring oscillator can be used [7].

2.4. Power feeding block

RFID tags can be categorized mainly into two i.e. active and passive tags, based on their source of power. Active tags are autonomous and carry their own power source usually in the form of an on-board battery. Active tags transmit a strong signal, and achieve a large read range. However, the presence of battery increases the size of the tag and they need to be replaced frequently, which is not suitable for the biomedical chip implanted in the body. Moreover, removing the tag from the body of living being for battery replacement is not also practical. Therefore, due to these constraints passive tags are preferred for biomedical applications. The passive tag does not have inbuilt battery, the power required for the tag is provided by the reader through magnetic coupling.

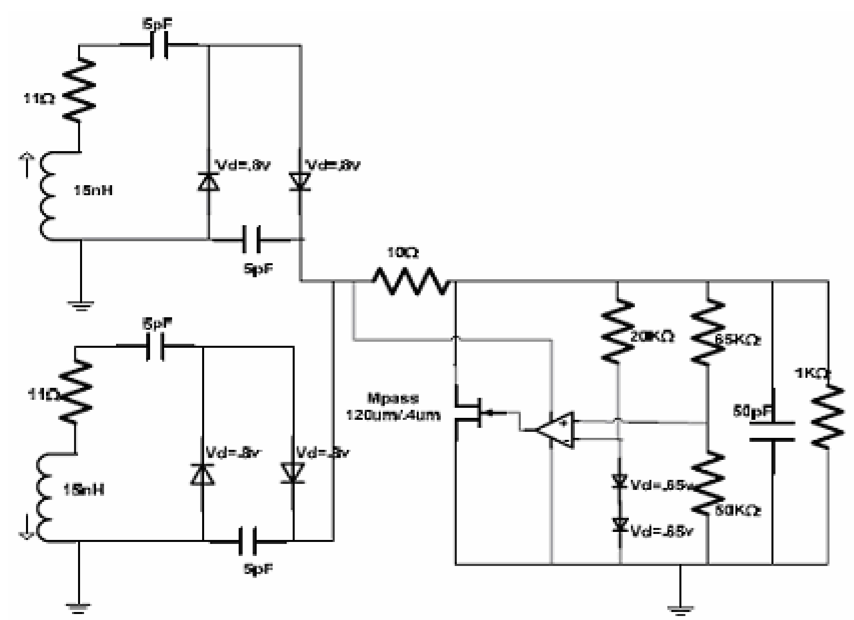

Fig.2. Power Feeding Block

The power feeding circuit is given in Fig.2.[5]. The function of the power feeding block is to convert the induced ac voltage into dc voltage. Charge pump circuit with diodes can be used for this purpose. The voltage across the load (i.e. tag circuit) is maintained constant even if there are variations in the induced voltage, with the help of regulator circuit. The output is sampled and given to error amplifier where the sampled signal is compared with the reference signal. The reference is generated with the help of diodes and a resistor. The error signal is fed back to the input of the MOS, which acts in such a way to compensate the voltage variations.

3. Sensor Modeling

There are numerous pressure sensors available using complex MEMS processes [9]. In this work, the main motivation behind the sensor design is to keep the fabrication steps minimum and therefore the design is capable of integrating on the VLSI RFID chip. Capacitive sensors are highly sensitive to absolute pressure and could be easily implemented using very thin diaphragm with a narrow capacitive gap and an air cavity. Capacitive sensors have high sensitivity but a small dynamic range because the gap between the capacitor plates must be very small to obtain a large capacitance. Silicon micromachining is used to fabricate the capacitive sensor presented in this paper. The silicon sensor has advantages of small size, low cost, and batch fabrication. In addition, since the silicon diaphragm has better mechanical properties, including freedom from creep, than metal diaphragms, good repeatability can be expected.

3.1 Structure and Principle

The conceptual geometry of the capacitive sensor consists of cantilever beam separated by air from a fixed ground plane. The cantilever beam in Fig. 3 can be viewed as semi infinitely long VLSI on chip interconnect separated from the ground plane by the dielectric medium (air).

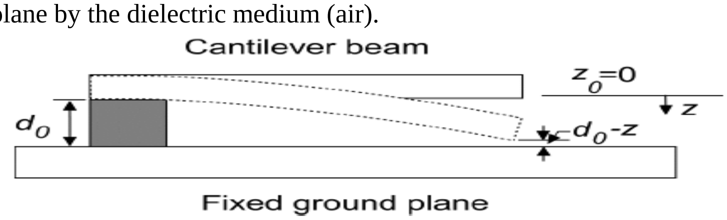

Fig. 3 Cantilever beam displaced due to pressure.

The fixed end is assumed to be at infinity where as the free end is finite. Under this assumption the fringing fields at the fixed end can be neglected. Thus the total capacitance between the cantilever beam and the ground plane is composed of the parallel plate capacitance, the fringing field capacitance due to the width of the beam, the fringing field capacitance due to the beam thickness, and the fringing field capacitance at the free end. The capacitance between the on chip interconnect separated from the substrate can be expressed as [10]

$$c = \in l\left[\left(w/do\right)+0.77+1.06\left(w/do\right)^{0.25}+1.06\left(h/do\right)^{0.5}\right]$$

Where l, w, h represent the interconnect length, width and thickness respectively $\in$ and d0 are the dielectric constant and the dielectric beam respectively. Effective value of d0 is the net initial distance between the plates minus displacement z.

3.2 Major Fabrication Steps

The fabrication sequence is shown in Fig. 4. The starting material is a 500µm thick typical silicon wafer.

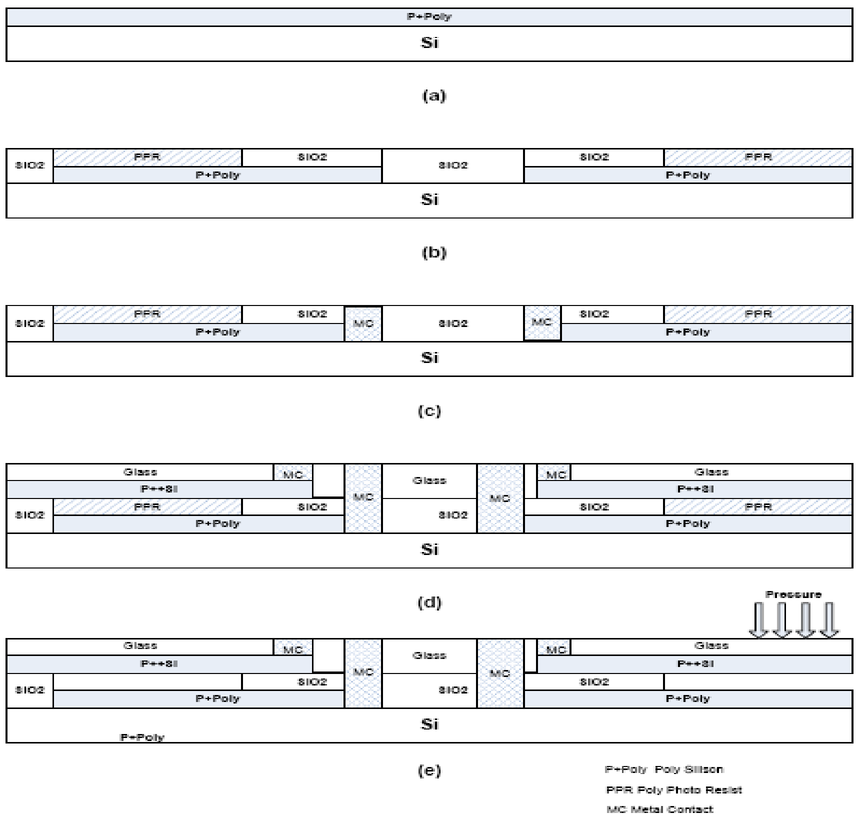

Fig . 4 . Fabrication process of sensor for heartbeat detection.

Heavily doped poly-silicon of thickness 10µm is deposited by chemical vapor deposition [CVD] on the top surface silicon wafer as shown in Fig. 4(a). Using a poly photo resist [PPR] mask, plasma etching is carried out selectively and SiO_2 layer of thickness 10µm is deposited by CVD as shown in Fig. 4(b). Small openings are provided in the SiO_2 layer for metal electrode contacts [MC] after etching of SiO_2 as shown in Fig. 4 (C).

Another layer of heavily doped poly-silicon of thickness 5µm is deposited on the top surface selectively and a glass layer is also deposited on the top surface of the heavily doped poly-silicon layer as shown in Fig. 4(d). Once again, windows are opened using lithography technique for aluminum electrodes. PPR is removed carefully, leaving behind the air gap in the final structure as shown in Fig. 4(e). Lead wires are attached with conductive epoxy.

The pair of poly-silicon electrodes on the left hand side form capacitance C1 and the other pair correspond to C2. Since both the plates on the left are fixed the capacitance C1 is constant. However C2 varies as pressure applied on the top surface changes.

3.4 Working Principle

The working principle of the sensor can be explained as follows. The capacitance increases with applied pressure because the movable silicon electrode moves toward the lower glass. The C/V converter circuit in Fig.5. will then transform the capacitance into an output voltage. The amplified voltage V_S is fed to the servo electrode, so that the electrostatic force F_e will increase and balance out the effect of the applied pressure. Therefore, the value of the servo voltage is thus related to the applied pressure P. Its dynamic range is very wide because the movable silicon electrode always stands in its initial position.

The electrostatic force per unit area is given by $F_e = \varepsilon_o V^2{}_s \big/ 2d^2$.

where the ε_o is the dielectric constant in vacuum and d is the space between the two electrodes of the servo capacitor C. When F_e is balanced with the pressure P from the above equation we get voltage $V = \sqrt{2d^2 P \big/ \varepsilon_o}$.

In the case of the fabricated device d= 5µm and voltage 1v corresponds to pressure of 1 mm Hg.

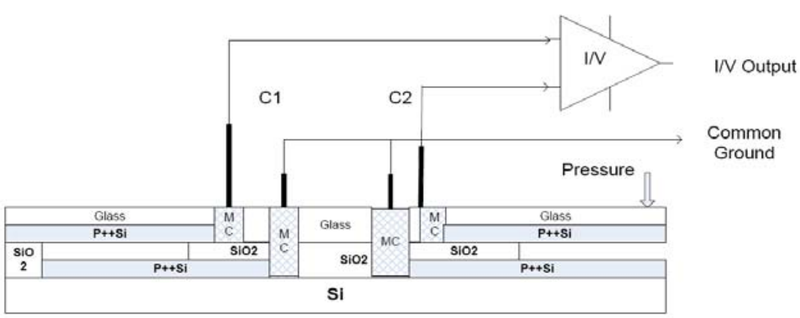

Fig.5. Structure of the Capacitive Pressure sensor.

3.3 Simulation results and Discussions

The entire analysis is carried out using Intellisute ThermoElectroMechanical simulator. The thickness of bottom plates of two capacitors is 10µm and the top plates are 5µm. The distance between the plates is 5µm and the protruding part of the capacitor C2 is almost equal to the part resting on SiO_2 base, to enhance adhesion and to prevent any break off due to pressure applied. The protruding part is approximately equal to 250µm. The ratio between the gap between the plates and the length of plates is chosen such that the aspect ratio is 50 which is suitable for practically removing the PPR to make the air gap. The length of plates of C1 is approximately the same. The width of the entire structure is around 200µm.

Impulse pressure of 1 mm Hg is applied uniformly on the top surface of the cantilever beam. Due to application of pressure on the top surface, the capacitance increases as shown in Fig.6. For a servo voltage of 1 V applied across the capacitor the initial value of the capacitance is 12 fF. Not much capacitance variation is seen for displacement up to 2µm. Beyond 2µm the capacitance varied drastically by a factor close to four to 48 fF. This is because for small displacements the charge gets redistributed resulting in little variance in capacitance. For large displacement due to electrostatic forces charge distribution is not uniform resulting in variation in capacitance. The capacitance returns quickly to its previous value of 12 fF which is not shown in the figure.

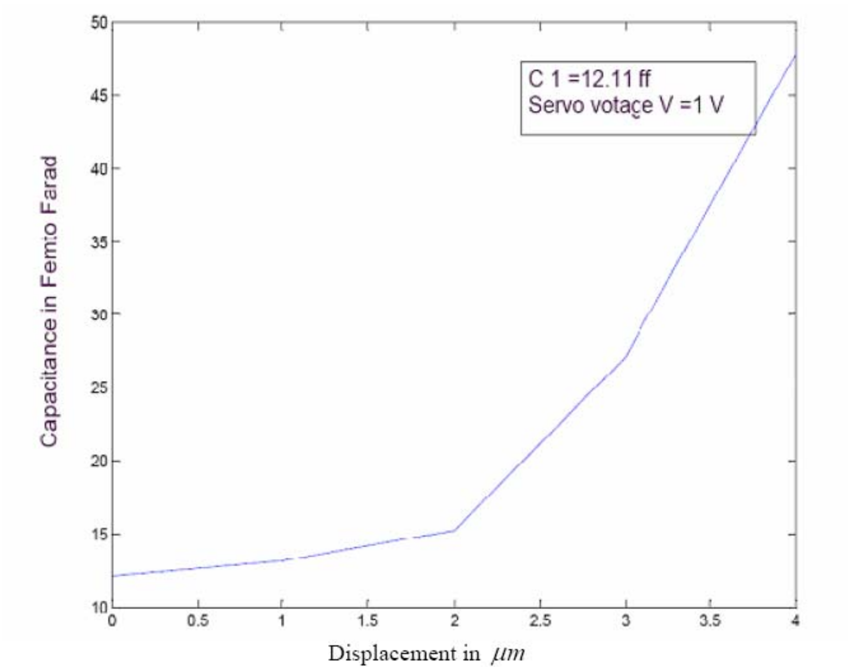

Fig. 6. Output plot of C2 with pressure/ displacement.

Since the impedance offered by the low value capacitors at low frequency is very high, a differential amplifier with low output impedance is required so that the signal current is effectively coupled to amplifier. A suitable differential amplifier circuit with shunt-shunt feedback connection is designed with C1 and C2 connected between the gates of the input MOS transistors and ground. Both the capacitors are charged using current sources and when there is a difference in the capacitor voltages the differential amplifier output will be switching from low to high. The capacitor C1 is fixed and is used as the reference capacitor. When no pressure is applied on plates of C2, differential output of the amplifier is low. When pressure/displacement is applied on the top surface of hanging beam, the capacitor C2 increases. This would lead to a higher differential voltage at the gate of MOS transistor to which C2 is connected and the output of the amplifier will be switched to high. Unwanted pressure variation other than the expected heartbeat variations also may slightly vary the capacitance C2.In order to nullify such effects, the threshold voltage which switches the output voltage from low to high is chosen to be twice the value of C1.This can be achieved by appropriately choosing the current levels and W/L ratios of the transistors. Hence when C2 increases by more than four times the value of C1 the amplifier output switches to high voltage.

4. Integration with the RFID Tag

The output from the differential amplifier circuit designed to switch on and off in accordance with the variations in the capacitance, is shown in Fig. 7.

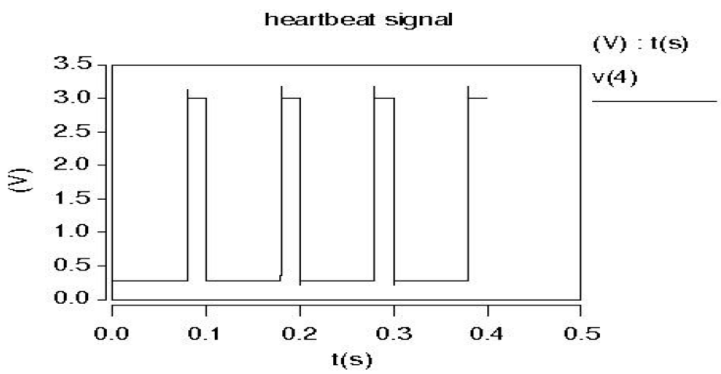

Fig. 7. Heartbeat Signal

The heart beat signal thus generated is modulated with a carrier wave of frequency 915 MHz generated from the power feeding block, shown in Fig.8.(a) The output DC voltage of the power feeding block is shown in Fig. 8.(b).The output of the modulation circuit is shown in Fig.9.(a). The output for the short duration is shown for clarity in Fig.9.(b). Entire circuit simulation is done in Hspice for 0.35µm CMOS technology.

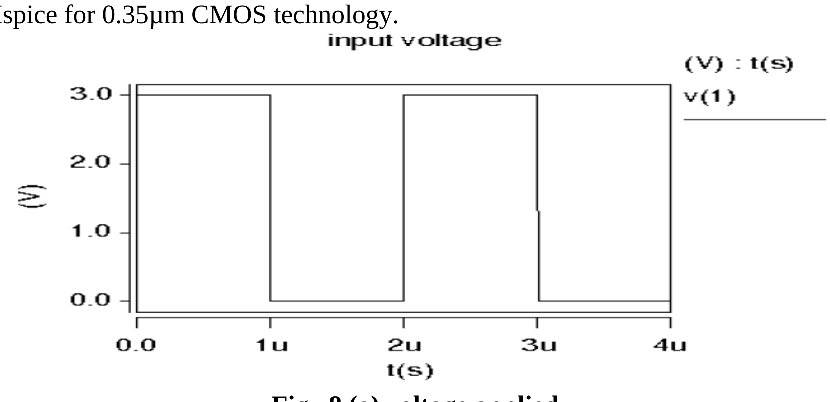

Fig . 8 (a) voltage applied

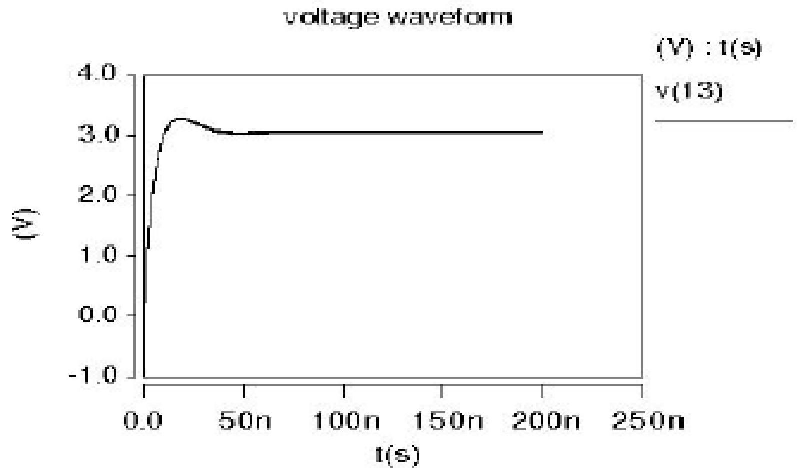

Fig 8 (b) Output voltage waveform of power feeding block.

The inductor of the power feeding block is optimized for maximum quality factor of 6 at 0.915 GHz, coupling coefficient of 0.3 and inductance of 15 nH, number of loops is 4 [5].

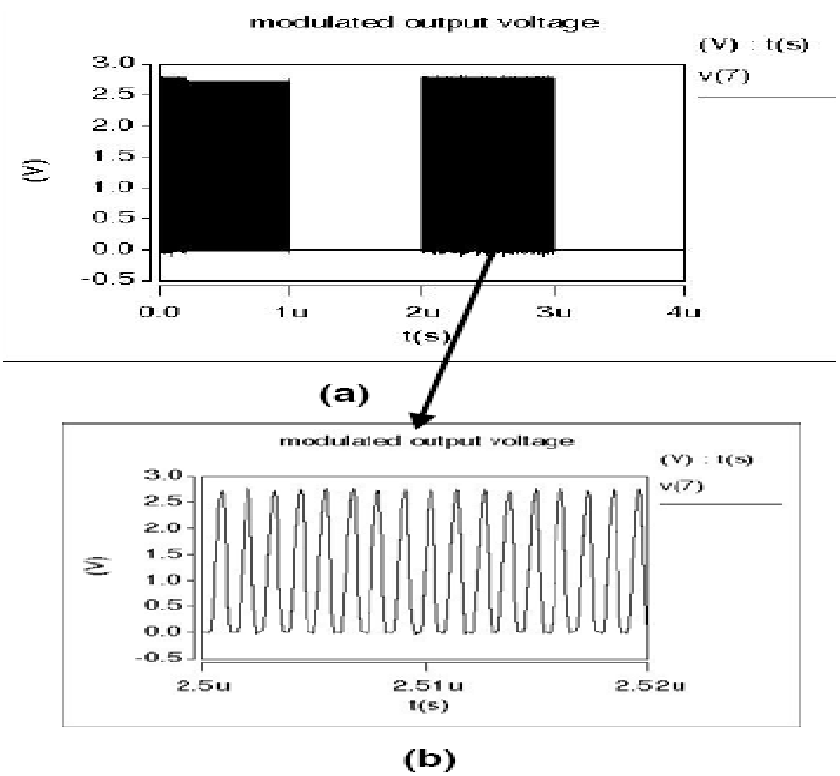

Fig .9. (a) Output voltage of modulated circuit (b) Output for short duration 2.5 usec to 2.52 usec shown for clarity.

5. Conclusion

A capacitive pressure sensor for detecting the heartbeat signal was discussed in this paper. A cantilever based capacitive pressure sensor for detecting the heartbeat signal was discussed. A simple sensor with only few process steps compatible to standard CMOS flow is proposed. The basic structure of the sensor along with fabrication steps has been studied. The feasibility of fabrication and immunity from unwanted pressure/noise variations are mainly considered during the optimizing of sensor. The final structure has an air gap dimension of 250µm X 200 µm X 5µm under a 5µm thick cantilever beam. The entire circuit is verified by simulation and the results obtained were satisfactory. This work would be very relevant for the sensing and remote monitoring of biomedical signals.

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