# FORCE MEASURING DEVICE APPLIED IN CARDIOPULMONARY RESUSCITATION

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**Abstract: This paper deals with a force measurement device that is applied in real and for training cardiopulmonary resuscitation. The measured force is shown on the computer monitor and the obtained data can be stored. To fulfill the requirement that it has to be flat and thin, a capacitive sensor is applied. This sensor is pulse driven to realize low power consumption, what is important when is used in the field. Also upper and lower part on the capacitor plates have to be adapted to the rescuers hand and patient sternum, where the sensor is applied. This sensor has a linear relationship between applied force and the output signal in the required range.**

## Introduction

In both types of cardiopulmonary resuscitation (CPR) (real and training) on-line monitoring of chest compression forces in time is most important. Force changes in time enable to determine also the compression frequency and the compression-relaxation ratio (duty cycle). Also compression depth can be estimated knowing chest compliance and the compression force.

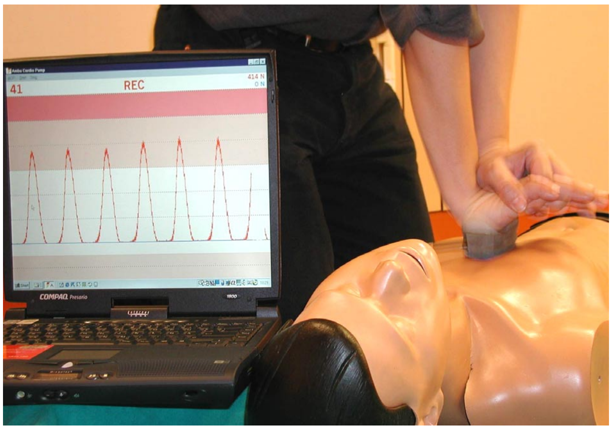


Figure 1. Resuscitation maneuver on a manikin with force measuring device. Compression patterns are shown.

The compression frequency is mostly in the range of 80-100 compressions per minute, and compression-relaxation ratio is approximately kept in the 1:1 ratio. The force changes during CPR can be on-line monitored and displayed on the computer monitor. Also the compression frequency and compression-relaxation ratio can be continuously calculated and displayed with numbers on the monitor screen. All measured data can be stored on a hard disc for later study, what is most important in rescuer training. For the rescuer training a manikin (puppet) is used [2]. This is shown in Fig. 1. To fulfill all these requirements a force measurement device with capacitive pressure sensor is described.

## Force transducer

For the force measurement in CPR most appropriate is found to be current pulse driven capacitive transducer. The short current pulses enable a very low current consumption of the transducer. Also pulse driven capacitive sensor, in comparison to a strain-gauge bridge powered with sine-wave voltage, requires only one parameter for zero adjustment. Furthermore, the capacitive sensor is easy to be made flat and thin, what is not possible to be done with a strain-gauge bridge.

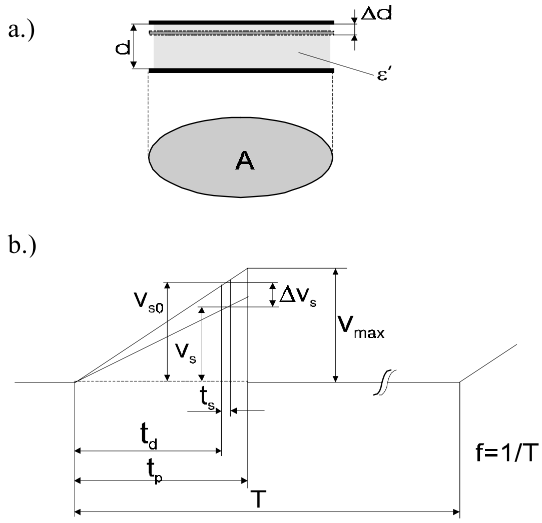


Figure 2. a) Capacitive sensor, b) Voltage on the capacitor during current pulse

The capacitive sensor has a capacitance $C_0$ given by relation

$$C_o = \frac{\varepsilon_0 \varepsilon' A}{d} \qquad (1)$$

where $A$ is the plate area, $\varepsilon'$ relative dielectric constant and $d$ is the thickness of dielectric as is shown in Fig. 2a. The dielectric is made by foamy material to be compressible, what means that thickness will be squeezed and changed by displacement $\Delta d$ when the force $F$ is applied according to the relationship between strain and stress

$$\frac{\Delta d}{d} = \frac{1}{E}\frac{F}{A} \qquad (2)$$

where the proportionality constant is reciprocal value of elastic-modulus $E$ (Young's modulus) or elasticity of dielectric material [3].

The voltage $v_s$ between the capacitive sensor's plates is defined by

$$v_s = \frac{I_c t_d}{C} \qquad (3)$$

When a constant current $I_c$ is applied during the time of pulse duration $t_p$ the voltage $v_s$ linearly increases and is limited by maximum permitted voltage $v_{max}$ of about 0,8 $V_{cc}$ (power supply voltage is $V_{cc}$). When the force $F$ is applied, during the current pulse duration $t_p$, the voltage at the time $t_d$ when sample is taken is

$$v_s = v_0 - \Delta v_s = \frac{I_c t_d d}{\varepsilon_0 \varepsilon' A}\left(1 - \frac{1}{E}\frac{F}{A}\right) \qquad (4)$$

It is evident from equ. (4) that voltage change $\Delta v_s$ is proportional to the applied force $F$

$$\Delta v_s = -\frac{I_c t_d d}{\varepsilon_0 \varepsilon' A^2 E} F \qquad (5)$$

This is shown in timing diagram in Fig 2b.

Yong's modulus $E$ is constant until applied force $F$ or stress $F/A$ of dielectric material reach a certain level, when $E$ starts to increase and the tendency to saturation occurs. If the stronger force $F$ should be applied, without saturation occurrence, capacitor plates area $A$ have to be enlarged or an other dielectric layer with higher elastic-modulus $E$ should be used.

**Circuit description**

In Fig. 3. block diagram of force transducer is given. The current source is realized with transistor Tr and it charges capacitive sensor $C_0$=120pF with the current $I_c$. The pulse duration $t_p$ (20 μs) is given by quasistable state of monostable M.M.1. Position $t_d$ (18 μs), where sample has to be taken is given by monostable M.M.2. and sample duration $t_s$ by monostable M.M.3. These monostables control the switches SW1 for pulse duration, and switch SW2 for sample duration $t_s$ (0,4 μs). The switch SW1 is at the same time part of the sample & hold circuit, which keeps the voltage of taken sample, to the next sample. Astable multivibrator A.M. is the clock working on frequency $f_{cl}$ = 10 kHz, so that the time $T$ between the pulses (pulse repetition time) is $T$ = 100 μs and duty cycle is 1:100. It means, that the average current is $I_{av}$ = $I_c$/100 and in this particular case for $I_c$= 10 μA is only 0,1 μA. In this way, with pulse driven capacitive sensor, power consumption is very low in comparison to resistive sensors.

After the sample & hold circuit, signal is amplified with amplifiers $A_2$ and $A_3$ to the required range, what is in this particular case at the output between 0 V – 4 V.

However, an additional problem here is "zero" voltage stability, or voltage $V_0$, which has been taken any time by manual zeroing when the force is not applied $F$=0N. This is solved by "automatic zeroing"

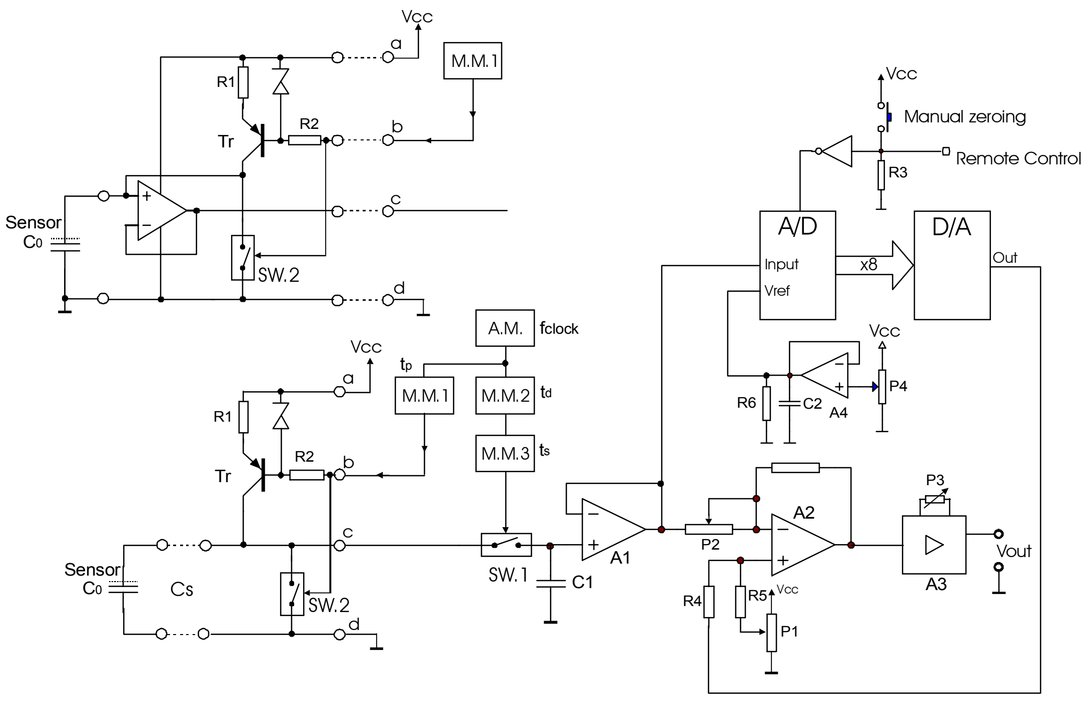


Fig. 3. Circuit diagram of capacitive force transducer (CFT)

with an A/D and D/A converter. The voltage $V_0$ at zero force is taken as a sample of short duration and brought to the input of A/D converter. The output of the A/D converter, given by 8-bit number, is kept until the "autozeroing" is applied again and converted by D/A converter in an analog voltage. This voltage is compared with the input voltage at the amplifier $A_2$. If they are adjusted to be equal, at the output of amplifier $A_2$ will be zero voltage. However it can be adjusted to give a small positive voltage, if is necessary or if the voltage at the input of the amplifier is below zero. The purpose of this "zeroing circuit" is to keep this small input voltage $V_0$ constant at the output of the device. The voltage range at the input of the A/D converter has to be diminished in this case by potentiometer $P_1$ because the voltage range of $V_0$ is small and about 1,5V.

The sensitivity of the force transducer defined as the $\Delta u_{max}/u$ for the maximum force, can be presented as the change of capacitance $\Delta C_{max}$ versus $C_0$, i.e. $\Delta C_{max}/C_0$. If the sensor is close to the electronic part of the unit, $C_0$ is unchanged. However if the sensor is connected over a cable, which has his own capacitance $C_s$, the sensitivity of the sensor will be reduced proportionally with the cable length, because now is $\Delta C_{max}/(C_0 + C_s)$ . This can be solved when the sensor is separated of the device with a voltage follower in the connection *c*, as is shown in Fig. 3. In this case the cable length has no influence on the sensor sensitivity. Therefore a small box with the amplifier should be connected close to the sensor. The connections of three wires with the cable screen *(d)* are denoted with the letters from *a* to *d* (Fig. 3).

Figure 4. View of capacitive sensor and electronic part of force transducer

In Fig. 4., a view of the sensor and the electronic part with cable is shown. The force sensor has on the both electrodes specially adapted plastic parts to the palm of rescuer hand and to the patient body at the compression point on the lower part of the sternum.

**Results**

In Fig. 5 transfer function of transducer is given as the relationship between the force *F* in N, applied on the sensor and the output voltage $V_{out}$ at the output of the transducer. It can be seen a slight hysteresis and a small nonlinearity.

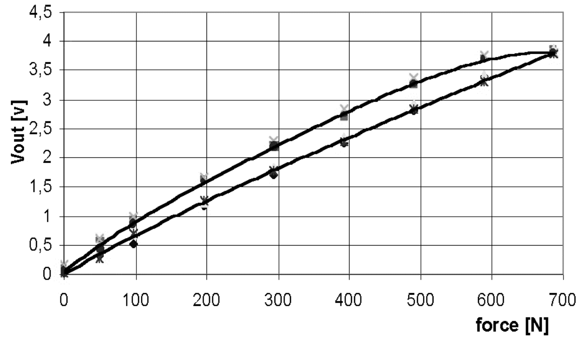


Figure 5. Output voltage $V_{out}$ versus force F of capacitive transducer

Calibration of the capacitive force transducer (CTF) was performed using a platform load cell (LC) arrangement as previously described. When applying a dynamic force pattern as used in standard cardiopulmonary resuscitation and comparing this signal with the signal obtained simultaneously with the LC, we noted that the CFT signal was delayed in comparison to the LC for approximately 45ms, as is shown in Fig. 6. During a real life resuscitation maneuver this lag time does not obscure the force measurement, as it remains unnoticeable for the rescuer. When the insignificant delay time of 45ms (versus period of about 800ms) is disregarded, an excellent agreement between two signals is obtained.

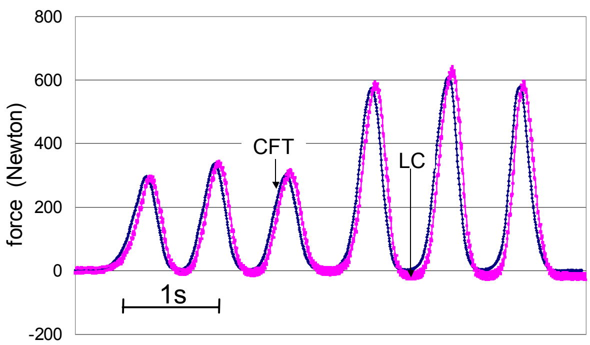


Figure 6. Simultaneous compression force measurement using the CFT and a platform LC with time delay 45ms

The correlation between CFT and LC during the compression phase (=upslope in Fig.6.) is shown in Fig. 7. Analysis according to Bland-Altmann (Fig.8.) yielded a small and insignificant difference between CFT and LC at forces smaller than 100 N.

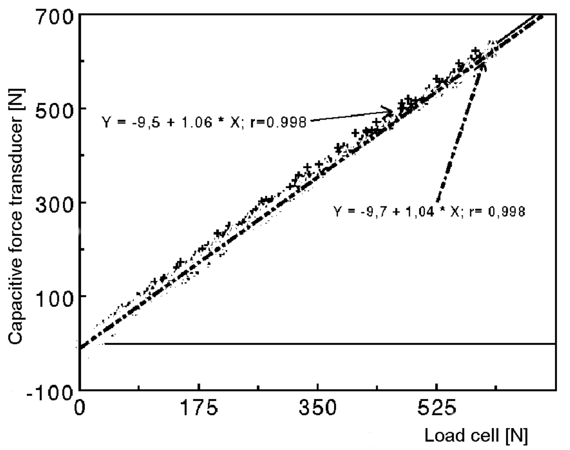


Figure. 7: Linear correlation between CFT and LC during the compression and release phase (data taken from Fig 6).

When correlating CFT and LC during the compression-release phase (downslope in Fig. 6) a small transducer hysteresis becomes evident from the linear regression coefficients (upslope: y= - 9,467 + 1,0576*x; downslope: y = -9,68 + 1,03978*x). However for the primary field of application – monitoring the performance of rescuers during training – hysteresis does in no way impair the analysis of the quality of the rescuers compression pattern.

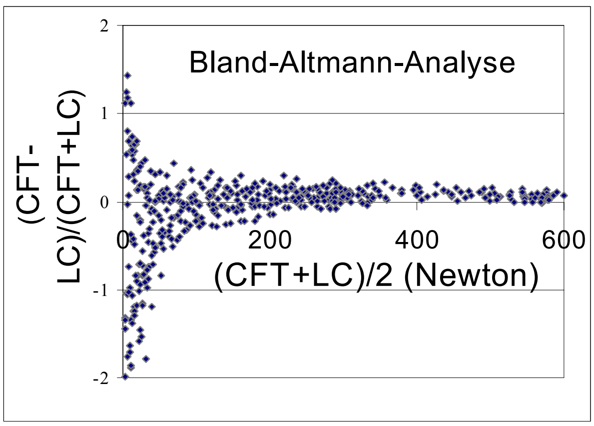


Figure 8. Bland-Altmann analysis of exerted force measured simultaneously by CFT and LC. Differences are very small at higher forces [4].

## Conclusion

The system for force measurement in real time during cardiopulmonary resuscitation (CPR) is described. The main advantage of this force measurement device is low power consumption to other systems with continuous current supply as is when the strain-gauges in deflection (Wheatstone) bridge are applied. Also when deflection bridge is used with one sensing element (the strain-gauge) the nonlinear relationship occurs. Low power consumption is particularly advantageous when movable system is required with laptop computer and for in field measurement as is in real CPR. Also the capacitive sensor can be made flat and thin what is very suitable to be placed between adequately shaped parts for the hand of the rescuer and the patient body. For a longer CPR, which may last a few hours, automatic zeroing is provided, because there is no time for zero adjustment during CPR process.

## Acknowledgment

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