

Actuation and Sensor Issues: Portable CPR Machine

L.Canan Dülger,
İzmir University of Economics
Mechanical Eng. Dept.
İzmir, Türkiye
canan.dulger@ieu.edu.tr

Melahat Kızıl
İzmir University of Economics,
Vocational School of Health
Services, İzmir, Türkiye
melahat.kizil@ieu.edu.tr

B. Karaaslan, T. Gülek, D. Eroğlu,
M.G. Güneş, A. Uzşen
İzmir University of Economics
Mechanical Eng. Dept.
İzmir, Türkiye
bkaraaslan.kb@gmail.com

Abstract— A portable CPR (cardiopulmonary resuscitation) B machine is designed as a prototype. A nonlinear viscoelastic model is used to consider the human thoracic properties. Based on the mechanical movements and force-acceleration accruals associated with CPR deployment. The system model is studied for the chest response to CPR by considering the viscoelastic properties of the rib cage and intrathoracic organs. It is then numerically simulated in MATLAB/Simulink. Actuation and sensor integration details are discussed. A DC motor is applied for driving the slider crank mechanism.

Keywords—CPR machine, Matlab-Simulink, Viscoelastic model, Chest model, slider crank mechanism, DC motors

I. INTRODUCTION

One of the leading causes of death is cardiac arrest. Those whose hearts have stopped may have a higher chance of survival using CPR. In an emergency, CPR is a first aid technique that can be immediately performed. The two most important factors, timing and the intervention quality, determine success in case of survival after cardiac arrest. Timing is the first, early identification and prompt treatment necessary. If the person's cardiac rhythm is detected, CPR or the use of a defibrillator should also be initiated. The Turkish Cardiology Association reports that there are 300,000 heart attacks and 125,000 heart attack-related deaths in Turkey annually. Between 20-30 % of heart attack victims die before they reach the hospital, while between 10-20% die subsequently. The death rate during a crisis's first 24 hours ranges from 30%-50% [1]. The American Heart Association and the European Resuscitation Council recommended that 30 chest compressions and two breaths should be administered initially during cardiopulmonary resuscitation (CPR). The chest should be checked at this point, and inflating it should take a few seconds [2]. Adult CPR requires chest compressions to be given at least 100–120 times per minute, and providing at least 5–6 cm deep [3].

CPR training is essential to provide immediate assistance during cardiac arrest. M. Kızıl has carried out a study on the effectiveness of CPR previously. First aid and basic life support have been performed at Izmir University of Economics. Seven different health programs were included with 153 volunteer students in 2017 [1]. The use of a training manikin is vital for CPR practice. Students can get hands-on practice by using such manikins. CPR training is given to project students from the Mechanical Engineering. Figure 1 shows the volunteer students

during CPR training in İEU Laboratories. CPR manikins are used during training. Two hands on the lower half of the sternum are used for adults, teenagers, and children. For infants, the same compression rate will be performed using only two fingers on the lower half of the sternum. This helped to improve the CPR prototype design and its construction details during the project.

Based on the load and motion properties needed for the mechanism, a motor is selected with the required torque and cycle rates. The motor has to generate chest compressions with the corresponding 5 cm deep (approximately) and 100–120 compressions per minute (cpm) were the choice when prioritizing for the selection process. This criterion optimization ensured compatibility between load characteristics, actuation components, and the need for a linear motion, making it valid for mechanical structure. Relevant sensors were included in the project to offer a basis for future system enhancements. Suitable sensors are purchased for pulse and chest compression force. Furthermore, preliminary mathematical modeling studies were performed to understand the system behavior. The responses to different parameter variations are acquired. The simulated model is analyzed. The requirement is for a system to work with CPR scenarios involving adults or teenagers. For the phases of the study (like sensor integration, control algorithms enhancement, mechanical structure, and the prototype manufacturing-3D printing).



Fig.1 CPR Training-İEU Laboratories

II BACKGROUND ON PROPOSED CPR

M. Kizil et al. [4] have presented a concept design based on active compression-decompression. Different piston-driven designs are then studied. L.C. Dülger et al. [5] have then offered a slider crank mechanism with detailed dimensions and simulations needed. A prototype has been printed in 3D, and its construction is completed. Figure 2(a) and (b) show the printed prototype using a slider crank mechanism with dimensions [5]. There have been numerous commercial attempts to address this requirement. The LUCAS™ and AutoPulse devices are examples of the two prominent types of MCPR systems on the market today [6]. M.J.Ghafoor et al. [7] and A. Athreya et al. [8] found that DC motors, particularly those that are gear-reduced, generate close human-like chest compressions when combined with slider-crank or cam-driven mechanisms. Kim et al. [9] demonstrated the feasibility of a system for CPR actuation that makes use of a 12V brushed DC motor and a 30:1 gearbox to produce the necessary torque.

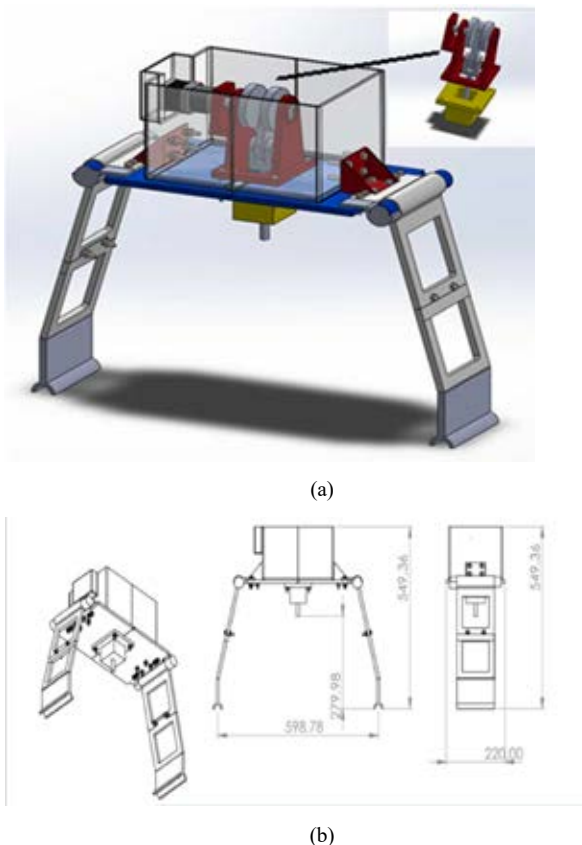


Fig.2 Referred Prototype -Slider crank mechanism with dimensions [5]

II. SENSORS FOR CPR DESIGN

The development of automatic cardiopulmonary resuscitation (CPR) systems has prioritized the use of embedded sensor technologies in recent years. Human mistakes, exhaustion, or inexperience are the leading causes of issues like inadequate pressure, improper depth, and irregular rhythms that are commonly seen in manual CPR applications. Sensor-supported technologies that provide real-time feedback are designed to improve the quality of CPR and guarantee safety [10]. One of the most widely utilized force-measuring sensors

in these systems is the load cell. By continuously and precisely measuring the force applied during compression, it aids in keeping the system within safe bounds. The measurements that load cells offer are crucial because too much pressure can injure the chest, and too little pressure can result in inefficient circulation. Ultrasonic sensors are used to measure the depth of chest compressions without contact. These sensors, which detect the distance between the piston and the chest surface with the help of sound waves, control whether the system provides the required 5–6 cm depth in each compression. [11]. Many CPR prototypes developed today use these three basic sensors together and process the data via a microcontroller. Sensor integration in CPR systems is a critical technology. Multi-sensor systems, where load cells, ultrasonic sensors, and accelerometers are used together, significantly increase the reliability and accuracy of CPR applications in a wide range from education to clinical settings. (Fig. 3)



Fig 3. CPR Sensors Used in Our Study

III. MATHEMATICAL MODEL-CHEST COMPRESSION

Design of CPR systems requires the approximation of thoracic behavior upon compressive load; otherwise, it will not correctly model the geometries and physiological dynamics in a medical compression. The goal is not simply to press on the chest but to simulate and deliver compressions in ways that mirror how the ribcage, sternum, lungs, and airflow systems interact during cardiac arrest scenarios. We review contributions that provide fundamental insights into the mechanical modeling of CPR that present data or methods to be used in our own nonlinear viscoelastic and energy-based formulations.

Ruiz de Gauna et al. [12] introduced a simplified viscoelastic model derived from real chest compressions collected via pre-hospital CPR monitors. Their work focuses on identifying mechanical parameters from actual resuscitation attempts. By computing chest displacement from force and acceleration data, each compression-recoil cycle was parameterized with an elastic stiffness constant k and separate damping coefficients for compression and recoil. J. Moradicheghamahi et al. [13] explored how thoracic geometry affects CPR outcomes using finite element simulations of anatomically variable ribcages. It then models the chest as a mass-spring-damper system. Kao et al. [14] have focused on CPR performance under controlled variations of thoracic stiffness. Their study used a multi-stiffness thorax manikin, in which subjects performed compressions, and the researchers determined depth, flow, and recoil.

A. Viscoelastic Modeling of Thoracic Mechanics for CPR

To accurately replicate the mechanical behavior of the human chest during cardiopulmonary resuscitation (CPR), we

introduced a dynamic model using a nonlinear viscoelastic model. This model is central to the design, simulation, and control of our portable CPR device, which uses a motor-driven slider-crank mechanism to generate cyclic compressions. Our goal was to simulate the thoracic response with sufficient detail to reflect stiffness, nonlinearity, energy loss, and real-time dynamics under the force and displacement cycles characteristic of CPR. A. Jalali et al [15] have performed a study to characterize the mechanical properties of a chest. The model implemented is built using Equation 1:

$$F = F_s + F_d + F_h + F_i \quad (1)$$

This total force F , acting on the sternum, is the total of elastic resistance (F_s), the viscous damping (F_d), the hysteresis damping (F_h), and the inertial response (F_i). Each of these terms reflects a distinct physical part that is important to the actual mechanics of CPR. They form a whole dynamic system that behaves realistically under simulation. The elastic force is defined as:

$$F_s = k_1 X + k_2 X^2 + k_3 X^3 + k_4 X^4 + k_5 X^5 \quad (2)$$

The fifth-order polynomial set above reflects the nonlinearity in thoracic stiffness. The lower-order terms k_1 and k_2 represent the initial compliance of the chest. The higher-order terms; k_3 , k_4 , and k_5 represent the geometric stiffness that occurs at greater depths. This is particularly important for compressions that are bigger than 3 cm. Because the chest will stiffen more strongly, both due to bone geometry and muscle tension. In the simulations, this equation produces realistic force curves and limits compression depth to the safe values. The viscous damping term is given by:

$$F_d = c_1 \dot{X} + c_2 \dot{X}^2 \quad (3)$$

Energy losses in the chest modeled by this equation depend on the speed. The linear damping, which is proportional to the compression velocity, is the first term that accounts for linear dissipation. The second one describes the nonlinear strain effects. This damping controls how smooth the actuator behaves. The hysteresis damping term is modeled as:

$$F_h = c_h \frac{\dot{X} \text{sgn}(\dot{X})}{\omega} \quad (4)$$

This term added internal energy dissipation in terms of not only velocity but also direction and frequency. The sign function makes sure that damping always takes place in both compression and release phases. The division by ω ensures that at faster rates, the damping weakens slightly, representing real biological hysteresis behavior. This lag is crucial to real-world CPR. It affects how much energy is transferred back into the chest and how well the blood flow is maintained over cycles. The inertial component is given by:

$$F_i = M \ddot{X} \quad (5)$$

Including this equation allows the simulation to correctly model force overshoot and lag during fast actuation. To generalize the model across patient sizes and simplify computational analysis, the model was non-dimensionalized [15]. Table 1 represents scale parameters. The variables were

scaled using $x = \frac{X}{X_0}$, $\tau = \frac{t}{T}$, $m = \frac{M}{M_0}$, the system's frequency was similarly transformed: $\Omega = \frac{\omega}{\omega_n}$, $\omega_n = \frac{2\pi}{T}$.

TABLE I PARAMETERS FOR THE CHEST MODEL

Parameter (Unit)	Value
X_0 (meter)	0,05
T (second)	0,06
M_0 (kg)	3,0

Thus the resulting dimensionless model is:

$$\begin{aligned} f &= f_s + f_d + f_h + f_i \\ f_s &= \alpha_1 x + \alpha_2 x^2 + \alpha_3 x^3 + \alpha_4 x^4 + \alpha_5 x^5 \\ f_d &= \beta_1 \dot{x} + \beta_2 \dot{x}^2 \\ f_h &= \frac{\beta_h \dot{x} \text{sgn}(\dot{x})}{\Omega} \quad f_i = \ddot{x} \end{aligned} \quad (6)$$

These equations were implemented in MATLAB to simulate the thoracic response under varying compression cycles. The model's coefficients were selected from validated research using porcine subjects due to their mechanical similarity to human chest geometry. The parameters used were [15]: $\alpha_1 = 1.51$, $\alpha_2 = -0.12$, $\alpha_3 = 12.38$, $\alpha_4 = -16.99$, $\alpha_5 = 10.94$, $\beta_1 = 0.08$, $\beta_2 = 1.15$, $\beta_h = 0.17$. The parameters were estimated for two periods (until 10 min. after CPR starting, and after 10 min. from CPR starting)

B. Simulation Results

In simulations, the force-compression depth plot produced a clear and physiologically realistic hysteresis loop that strongly resembled the patterns seen in actual CPR data. As compression depth increased, the model generated a nonlinearly rising force curve, while the release path followed a lower-energy trajectory. The graph is beneficial as changes can be made in the crank radius, linkage ratio, and compression limits in the slider-crank mechanism used. (Fig.4). We then calculated the width and vertical amplitude of the loop to verify that our mechanical output would fall within the clinically accepted range of 5–6 cm at around 400 N maximum force. Any deviations beyond that would require redesigning the mechanical limits or motor torque. The plot also helped us to confirm that our system would not introduce sharp force spikes or unsafe recoil behavior during release, an essential consideration for rib safety.

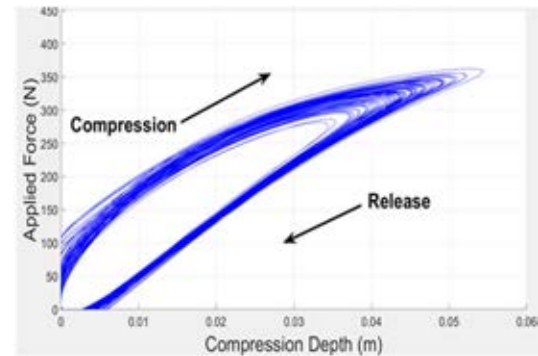


Fig. 4 - Simulated Force vs. Compression Depth -Thoracic Stiffness Behavior

IV. ACTUATION DETAILS FOR CPR SYSTEM

One of the initial and most critical steps in the actuation system design of our portable CPR machine was determining the most suitable motor type that could reliably generate repetitive, high-force compressions in line with AHA guidelines [3]. Brushed DC motors are the ideal compromise between high torque output at low voltage, PWM speed control compatibility, and high-volume/low-cost. A 12V brushed DC motor of a 30:1 planetary gearbox is used. This motor has an approximate no-load speed of 1300 RPM (13-14 Nm at output after gear reduction). The mechanism will provide mechanical leverage and control through the compression phase, which could resemble the proper 5-6 cm compression depth for CPR. The motor is integrated into the mechanical frame, with a 3D-printed coupling linking to the compression unit. The entire system is controlled by means (force load cell, ultrasonic sensor for depth) and processed in real time by the Arduino Due. These features allow for the editing of energy delivery during compression depth, rate, and force to prevent over-pressure and keep optimal performance [16-17]. Figure 5 shows side and top view photos from the designed prototype. Here, the motor and power system are activated. Sensor box design is completed, but not completely activated.



Fig.5. Designed Prototype with Actuation and Sensors

V. CONCLUSIONS

In this work, the development of a CPR (Cardiopulmonary Resuscitation) device has been nearly completed. A control hardware control was developed to supervise the motor's actuation. The system viscoelastic modeling is performed using MATLAB/Simulink. The sensors, an accelerometer, an HCSR-04 Ultrasonic sensor, and a load cell were chosen. Nevertheless, these sensors are not yet interconnected; however, full integration is planned in the subsequent stages of the project. Currently, the system can operate manually at an adjustable speed. It performs basic compression motions in the required range.

A nonlinear viscoelastic thoracic model is applied. The nonlinear viscoelastic model was also able to describe well the complex mechanical response of the chest wall to repeated compressions. The model enabled us to confirm that the force output fell below the 500 N safety limit while maintaining AHA-compliant compression depths of 4-6 cm. The results verified that a 12V brushed DC motor with a 30:1 planetary gearbox delivers adequate torque for safe and effective

compression, and that the crank-slider configuration translates this force into stable cyclic motion. It can be an affordable model with sensors for CPR training.

ACKNOWLEDGMENT

The study was conducted by Izmir University of Economics, Faculty of Engineering. It was carried out within the framework of FENG 498 projects between Dept. of Mech. Eng. and Vocational School of Health Services-Paramedics. We also thank I. Kulali for his contribution to viscoelastic modelling.

REFERENCES

- [1] M. Kızıl, Determining the effectiveness of Cardiopulmonary Resuscitation using a manikin CPR, December 2022, Hastane Öncesi Dergisi, 7(3), DOI: 10.54409/hod.1150760, pp.303-316.
- [2] D.A. Zideman, EDJ De Buck, EM Singletary, P. Cassan, A.F. Chalkias, TR Evans, C.M. Hafner, P.G. Vandekerckhove, European Resuscitation Council Guidelines for Resuscitation 2015, S.9, First aid Resuscitation, 95, 278-287
- [3] RW Neumar, M. Shuster, CW Callaway, LM Gent, DL Atkins, F. Bhanji, SC Brooks SC, AR de Caen, MW Donnino, JM Ferrer, ME Kleinman, SL Kronick, EJ Lavonas, MS Link, ME Mancini, LJ Morrison, RE O'Connor, RA Samson, SM Schexnayder, EM Singletary, EH Sinz, AH Travers, MH Wyckoff, MF Hazinski, Part 1: Ex.Summary:2015 American Heart Ass. Guidelines, Cardiopulmonary Resuscitation and Em. Cardiovascular Care, Circulation. 2015, Nov 3; 132, pp. 315-67.
- [4] M.Kızıl and L.C.Dülger, Design of a Device for CPR (Cardiopulmonary Resuscitation, Tıptekno 2022, 31 Ekim-2 Kasım 2022, Belek-Antalya.
- [5] L.C.Dülger, M.Kızıl, M.Atçeter, D.D.Demircioğlu, D.Yiğit, B.Güneş Design and Construction of a Portable CPR Machine, Tıptekno 2024, 10-12 Ekim 2024, Gümüşlük/Bodrum-Muğla.
- [6] <https://www.lucas-cpr.com>, Assessed 03/08/2024
- [7] M. J. Ghafoor, M. Mujeeb-U-Rahman, M. Jamal, and A. Ahmed, "Low-Cost and Portable Cardiopulmonary Resuscitation Machine," Dept. of El. Eng., Information Technology University, Lahore, Pakistan,
- [8] A. Athreya, S. Gupta, and R. Bansal, "Design and Development of an IoT-Based CPR Delivery System," J. Med. Devices Eng., V45, no. 3, pp. 210-218, 2023.
- [9] J. Kim, Y. Park, and T. Seo, "Low-Voltage CPR Actuation Mechanism Using Brushed DC Motors," IEEE Trans. Biomed. Circuits Syst., V13, N. 2, pp. 344-351, 2019.
- [10] Xie, X., Wu, X., and Liu, Y. Design and Evaluation of CPR Emergency Equipment for Non-Professionals. Sensors, 23(13), 5948. MDPI, 2023
- [11] G.Rugumira, ME Mkiramweni, N.T. Msangi, "Design and Implementation of Cardiopulmonary Resuscitation Simulation Model, Tanz. J. Eng. Tech. J., V41-No.4, Dec 2022, pp 133-147.
- [12] S. Ruiz de Gauna, JJ. Gutiérrez, CL. Sandoval, JK. Russell, I. Azcarate I, JA Urigüen, DM González-Otero, MR. Daya, "Characterization of mechanical properties of adult chests during pre-hospital manual chest compressions through a simple viscoelastic model". Comput Methods Programs Biomed. 2023 Dec; 242:107847. doi: 10.1016/j.cmpb.2023.107847. Epub 2023 Oct 10. PMID: 37852146.
- [13] J. Moradicheghamahi, G. Fortuny, JM. López, D. Puigjaner, J. Herrero, Y. Azeli. The effect of thoracic dimensions on compression depth during cardiopulmonary resuscitation. Int J Numerical Methods Biomed Eng. 2023, Jul;39 (7):e3718. doi: 10.1002/cnm.3718. Epub 2023-May 11. PMID: 37166223.
- [14] Chia-Lung Kao, Jui-Yi Tsou, Ming-Yuan Hong, Chih-Jan Chang, Fong-Chin Su, Chih-Hsien Chi, "Effect of thoracic stiffness on chest compression performance- A prospective randomized crossover observational manikin study", Heliyon, V8, I10, 2022, e10990, ISSN 2405-8440.
- [15] A. Jalali, V. M. Nadkarni and C. Nataraj, "Modeling mechanical properties of the chest during the cardiopulmonary resuscitation procedure," Computing in Cardiology 2014, Cambridge, MA, USA, 2014, pp. 13-16.
- [16] Y. Xie, L.Zhang, and W. Chen, "Sensor-guided Feedback Systems in CPR Training and Application," Sensors, vol. 23, no. 5, pp. 2244-2255, 2023.
- [17] T. Iwata, K. Yamashita, and M. Nishimura, "Development of chest compression monitoring system using accelerometer". J. of the Japanese Society of Intensive Care Medicine, 2013, 20(3), 179-183.