ASPECTS OF THORAX BIOMECHANICS AND THEIR IMPLICATIONS IN IMPROVING CARDIOPULMONARY RESUSCITATION TECHNIQUES

M.A. UDRIŞTE, A. NEAGU***

*County Hospital, 36, Aleea Spitalului, 110283 Piteşti, Romania,
e-mail: udriste_mihai@yahoo.com
**Center for Modeling Biological Systems and Data Analysis, Department of Biophysics and Medical Informatics, "Victor Babeş" University of Medicine and Pharmaceutics, Timişoara, 2, E. Murgu Square, 300041 Timişoara, Romania

Abstract. In accordance with the new cardiopulmonary resuscitation (CPR) guidelines, which focus on infusion rather than on ventilation, a hands-only method was proposed that improves both ventilation and perfusion by using an elastic support placed between the patient's back (along the thoracic vertebrae T8-T10) and the resuscitation plane, thus lifting the spine above the ground. This method has been called resuscitation on support or suspended resuscitation. Using numerical simulation techniques based on finite element analysis, this paper proves that the proposed elevation causes an increase in the anterior-posterior diameter of the thoracic cavity. After each compression during CPR, the elastic support brings the spine back to its initial elevation. This study proves that, provided that the airways are permeable, the elastic support causes an increase in the volume of passively inspired air.

Key words: myocardial infarction, finite element analysis, ANSYS, ventilation.

INTRODUCTION

Every year throughout the world hundreds of thousands of people experience myocardial infarction. Statistics show that in Europe approximately 275,000 people suffer a myocardial infarction every year [6], and in the United States of America their number is 400,000 [3]. Most cases of myocardial infarction occur out of hospitals, being followed by short life expectancy. During a stroke, the heart does not function as a proper pump. Therefore, a prompt and efficient cardiopulmonary resuscitation (CPR) is vital for the patient [7].

Over time the methods of CPR have been studied extensively, but significant progress occurred a few years ago, in 2005, when the American Heart Association (AHA) and other medical organizations have suggested a change in the way CPR is

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performed [1]. The new guidelines suggest a hands-only approach to CPR; the rescuer is not supposed to apply a mouth-to-mouth resuscitation. According to the new guidelines, chest compressions should be done at a rate of 100 compressions per minute and the chest should be compressed by a full 1.5 to 2 inches, eliminating pauses for ventilation. The guidelines urge that focus be shifted on heart compressions (perfusion) rather than on ventilation. The new AHA standards call for square-wave compressions and emphasize the importance of delivering compressions of adequate force, depth and frequency [1].

Taking into account spine biomechanics, this work aims to improve the CPR technique. In the context of the novel AHA guidelines [1], the present computational study brings quantitative arguments for the efficiency of a novel technique proposed by our group [9–11].

The new technique involves the use of a support placed between the flat hard plane on which the patient is lying and the patient's back; the support is placed along the spinal column, facing the 8th to10th thoracic vertebrae (Figs. 1, 2). This method was presented for the first time in 2004 [9] and then at the MEDIS-2006 meeting [10]. The advantages of this method, understood until now, refer to improved perfusion due to a better compliance of the rib cage and, thereby, a more efficient compression of the heart. This was highlighted by a biomechanical study of the thoracic cage, focusing on (i) the distance between the lower third of the sternum and vertebrae T8-T10, and (ii) the stress distribution in the ribs [11].

This paper asks the question whether the new method also improves ventilation during CPR. To answer this question, we performed an *in silico* biomechanical study of the kinematic chain consisting of vertebrae in direct correlation with the movement and deformation of the rib cage.

Given the special nature of CPR, every second is vital, and experiments are hampered by ethical issues. Thus, our method of choice was a computer simulation in a widely known, commercially available software environment, Ansys (ANSYS, Inc., USA), using a well-established numerical technique, the finite element analysis [2]. By choosing a standard numerical technique, we aimed to make our study reliable and reproducible.

The novelty of the present approach consists in the way the problem is formulated and how the mechanical parameters of the rib cage are extracted from the literature. The limitations of the method stem from the difficulty of building a mechanical model that captures the full complexity of the human body. The viscoelastic parameters of living tissues represent an open, vast field of biomechanical research. The stress-strain relationships (constitutive equations) are not known for all thorax components. Nevertheless, we think that the average values of the material constants considered here represent a good starting point for quantitative biomechanical studies of the thoracic cage.

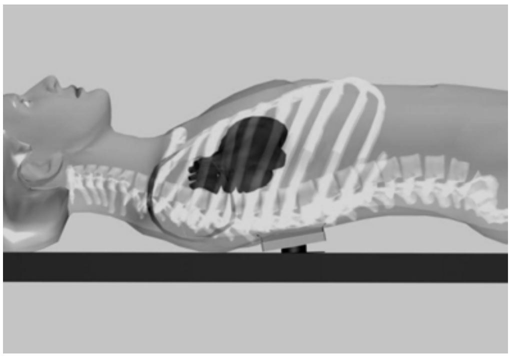

Fig. 1. Suspended resuscitation – lateral view. Illustration of positioning the support on which resuscitation is performed.

METHODS

Systems of high complexity are customarily studied by means of modeling and numerical simulation. Analyzing cardiopulmonary resuscitation on the basis of a virtual model has been considered the most appropriate for the present study. In this way new procedures can be designed and evaluated.

Our study of thorax deformations during CPR is based on computational biomechanics. We had in view an accurate model geometry for a high-fidelity representation of the kinematic chain made of bones and inter-vertebral joints.

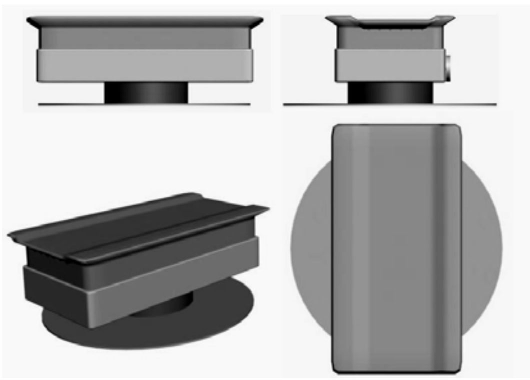

Fig. 2. Lateral, frontal, upward and long distance view of the support designed for suspended resuscitation.

The mechanical properties of soft tissues were not included in the simulation. In terms of geometry, the model comprises ribs, sternum, thoracic vertebrae, inter-vertebral discs, the lungs and the heart, together with the large emerging blood vessels (Fig. 3).

A computer assisted design (CAD) software was used for graphic editing. The prototypes obtained from medical imaging by assembling two-dimensional images proved to be very useful in setting up the three-dimensional model. The spatial geometric model of the thorax (Fig. 3) comprises 82 parts, 477,573 nodes and 260,738 elements.

The displacements thorax components during CPR have been studied by means of a biomechanical analysis using the finite element method, which is one of the most common methods used in mechanics for analyzing stress and deformation.

To define the mechanical properties of the ensemble, we used the following material parameters [5, 6, 8]: (i) a Young modulus $E = 11$ GPa and a Poisson's ratio $\nu =$ 0.35 for spine bones, (ii) $E = 13$ GPa and $\nu = 0.35$ for ribs and stern, and (iii) $E =$ 200 MPa and $\nu = 0.4$ for cartilage.

The stages of the computational procedure comprised importing the geometry, defining the materials, meshing the model, defining the supporting and loading points, running the simulation, and obtaining the solution.

As points of support, we considered the upper surface of the first cervical vertebra and sacrum bone surfaces corresponding to the sacroiliac joint. The loading points were the spinous processes of vertebrae T8-T10 (more precisely, their peaks).

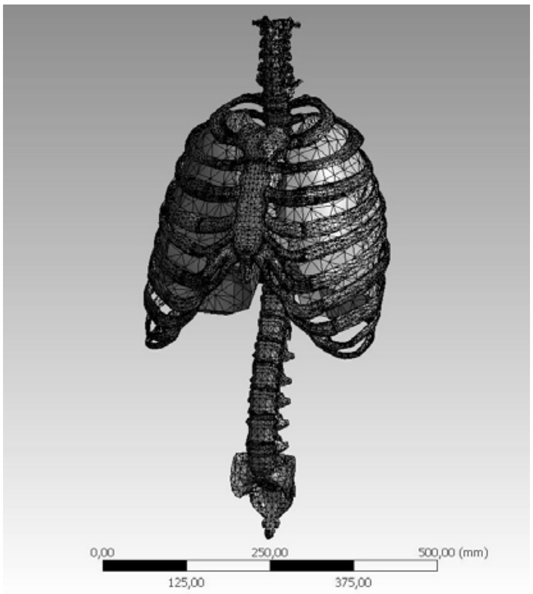

Fig. 3. The three-dimensional model of the thoracic cavity that includes the heart and lungs.

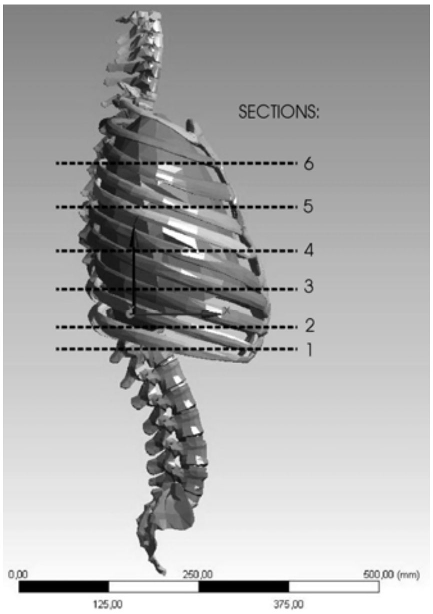

Fig. 4. Cross sections through the rib cage that were considered in the study of thorax deformations.

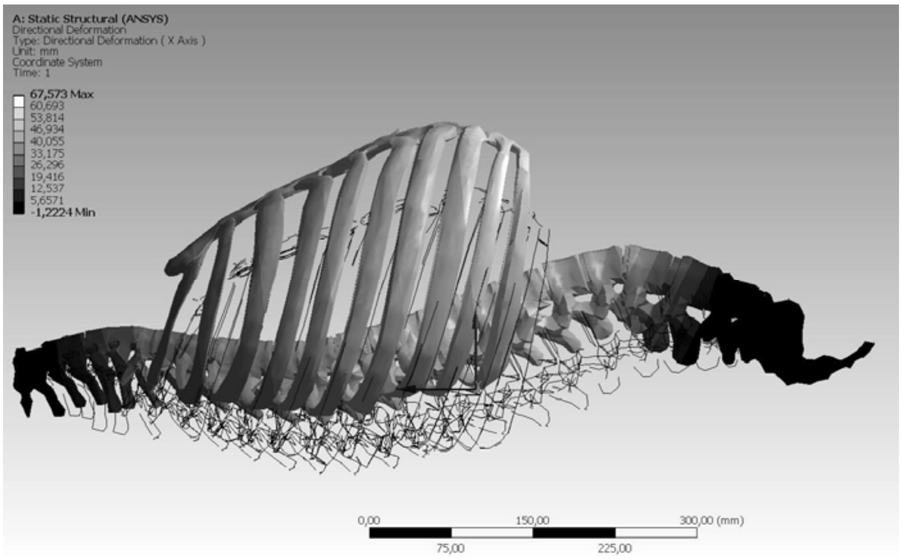

Fig. 5. Shape changes of thorax components due to a 60 mm displacement of the loading points (spinous processes of vertebrae T8-T10) along the *Ox* axis (from bottom to top). The gray-scale picture shows the final state of the model, whereas the black wireframe represents the initial state, associated to the patient lying on the resuscitation (*yOz*) plane. Shades of gray represent the magnitude of *Ox*-displacements of the components according to the numerical scale shown on the left side of the figure.

All simulations start from the same initial state that corresponds to the patient lying on the resuscitation plane (a horizontal, flat, hard surface that supports the patient during CPR). We associate a rectangular coordinate system to the model such that the resuscitation plane is the *yOz* plane with the *Oz* (*Oy*) axis parallel (perpendicular) to the spine.

In the first set of simulations, a displacement of 30 mm was imposed at the loading points along the posterior-anterior direction (associated to the *Ox* axis). In the second set of simulations, a displacement of 60 mm was imposed at the loading points. The displacements of 30 mm and 60 mm, respectively, mimic spine elevations caused by an elastic support of the given thickness (Fig. 1).

In order to point out the role played by the support during a CPR, we computed the shape changes of the thorax caused by the given spine elevation. More precisely, in the *xOy* plane, we defined six cross sections through the rib cage (Fig. 4), and, in each cross section, we calculated the *Ox* displacements of the anterior parts (stern, anterior costal margin) and of the posterior parts of the thorax (vertebrae).

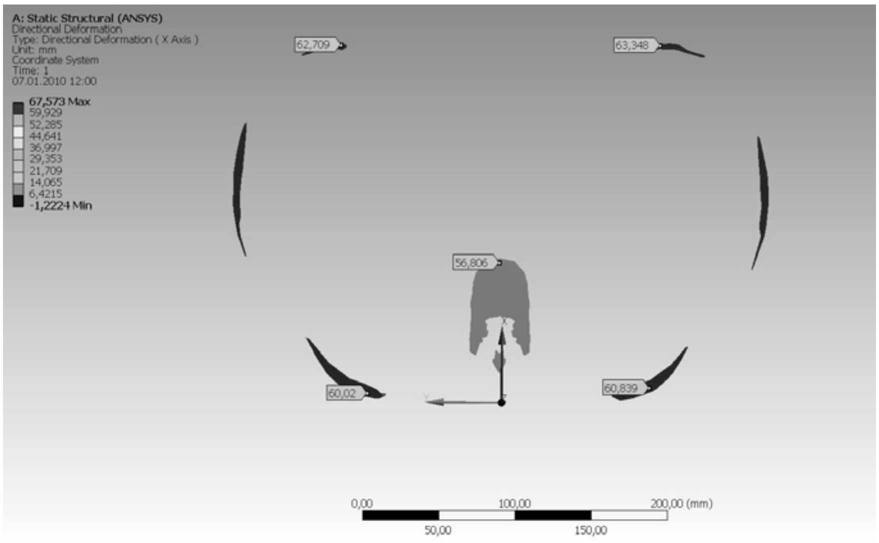

Fig. 6. Displacements of various components in the second plane of cross section (defined in Fig. 4). Numerical values give the *Ox* – displacements (in mm) of sample points on the ribs (top) and vertebrae (bottom). The deformation was triggered by a 60 mm displacement of the loading points.

The displacements of thorax components were computed using Ansys (ANSYS, Inc., SUA), a commercially available finite element analysis (FEA) software, on a personal computer hardware platform with a 3.2 GHz Intel dual-core processor. Simulations had durations of the order of minutes.

RESULTS AND DISCUSSION

The FEA yields the extent of deformation of an elastic body as a result of a well-defined change in the boundary conditions. Here we are interested in shape changes of the thoracic cage that occurs due to lifting the backbone along the peaks of the spinous processes of vertebrae T8-T10 (loading points). With the aim of optimizing cardiopulmonary resuscitation, we followed the geometrical parameters that indicate changes in thorax volume.

The results of a representative simulation are depicted in Fig. 5, a lateral view of the model in its initial state (wire frame) and in its final state (gray-scale image). In the latter image, slightly displaced regions are shown dark gray, whereas strongly displaced regions are light gray. The movement of the loading points was imposed to be 60 mm along the vertical, *Ox* direction. As a result, the thoracic and lumbar curvatures change, causing an overall deformation of the thorax.

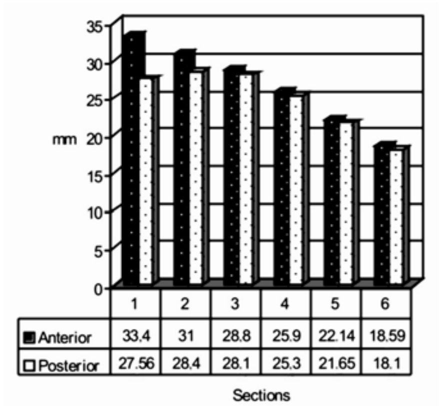

Fig. 7. Displacements of the anterior components of the thorax (stern, anterior costal margin) compared with displacements of the posterior components (vertebrae) in the cross sections shown on Fig. 4. Thorax deformation resulted from a 30 mm displacement of the loading points.

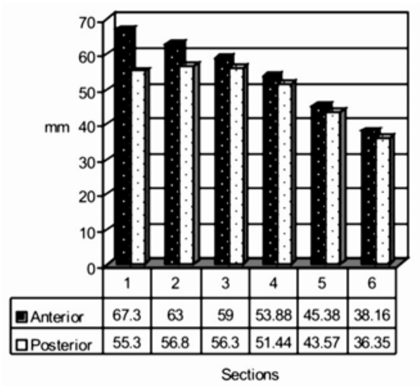

Fig. 8. Displacements of the anterior components of the thoracic cage compared with the displacements of the posterior components in the cross sections defined in Fig. 4. Thorax deformation resulted from a 60 mm displacement of the loading points.

As shown on Fig. 6, the various components from the 2nd cross-sectional plane (defined in Fig. 4) suffered different displacements due to the overall deformation of the system. More specifically, the frontal surface of the thoracic cage moved more than the vertebrae along the posterior-anterior (Ox) axis.

As a quantitative example, in the plane of the first cross section the displacement of the anterior costal margin was 33.4 mm, while displacement of the corresponding vertebra was only 27.56 mm. Their difference, of about 6 mm, is the amount by which the anterior-posterior diameter of the rib cage increased at the first cross section. This observation is true for all cross sections (Figs. 7 and 8).

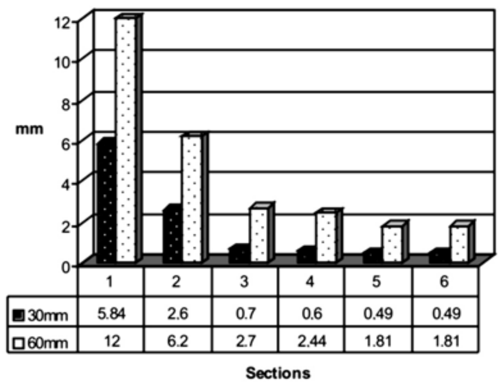

Fig. 9. Increments of the anterior-posterior diameter of the thoracic cage as a result of 30 mm (black), or 60 mm (white) lifting of the backbone at the spinous processes of vertebrae T8-T10.

Figure 9 shows the extent by which the anterior-posterior diameter of the thorax increases while the spine is displaced upwards by 30 mm or by 60 mm at the level of vertebrae T8-T10. The larger the lifting of the loading points, the larger is the increase in the anterior-posterior diameter of the thoracic cage. This corresponds to an increase in the thorax volume. The more pronounced expansion of the thorax is observed in its lower regions, near the abdomen.

Nevertheless, the lifting of the loading points should not be too large. An estimation of the largest acceptable elevation goes beyond the objectives of this study; it would require a more refined model, with precise constitutive equations for the inter-vertebral cartilage, and with soft tissues included. The range of elevations investigated in the present work (30–60 mm) was chosen for an average adult on the basis of whole-body biomechanics [4].

To test the influence of material constants on the above results, we conducted a series of simulations with different mechanical properties associated to the components of the model system. For example, by associating an elastic modulus of 13 GPa and a Poisson's ratio of 0.35 to all bone parts, we observed that the displacements of thorax components changed by less than 10% in comparison to the displacements presented here. In simulations with mechanical properties of cartilage (an elastic modulus of 200 MPa and Poisson's ratio of 0.4) associated to all hard tissues, we obtained an 11 mm change in the posterior-anterior diameter of the thorax in the first plane of cross section for a 60 mm lifting of the spine at the loading points (results not shown).

CONCLUSIONS

We performed a biomechanical study of a novel cardiopulmonary resuscitation technique. Based on the finite element method, our analysis focused on the dynamics of the whole thoracic cage and of the spine.

According to our results, in all cross sections defined in Fig. 4, the displacements of the sternum and of the anterior costal margin along the anterior-posterior (Ox) axis have exceeded the corresponding displacements of the vertebrae (Fig. 9). This implies that, by lifting the spine along vertebrae T8-T10, the anterior-posterior diameter of the rib cage increased (Figs. 7 and 8). This amounts to an increase in the thoracic volume.

Provided that the airways are permeable, the CPR technique suggested in this work leads to an increase in the amount of passively inspired air. To yield this benefit throughout resuscitation, the support of the spine (Fig. 1) must be compressible and flexible, assuring that the spine returns to the same elevation after each cardiac compression.

Our model is unable to capture the full mechanical complexity of the human body. Due to inherent simplifications (such as the lack of soft tissues and ligaments, some truncations of the geometry, and the lack of degrees of freedom for certain components) it is expected that the numerical results do not describe exactly the clinical context. Nevertheless, the essential conclusion, namely that thorax volume increases due to spine lifting, seems to be correct, since it holds for a wide range of model parameters.

Since the new cardiopulmonary resuscitation protocols favour ventilation by compression over mouth-to-mouth resuscitation, we think that suspended resuscitation can bring an important benefit by increasing the efficiency of ventilation.

Acknowledgments. We dedicate this work to the memory of the late Professor Iosif I. Nagy, who shaped our ideas about this topic.

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