Abdominal Finite Element Model for Traffic Accidents Injury Analysis

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ABSTRACT: The abdominal cavity is a vulnerable region of the human body. The severe and critical abdominal injuries arising during traffic accidents are the second most important bodily injuries after thoracic injuries. The injuries are predominantly caused by contact with the lap belt, steering wheel, armrest, dashboard, side door, etc. However in terms of investigating the biomechanical response of the abdomen, experimental studies have turned out to be particularly difficult to perform and the results obtained are not easy to interpret. Therefore attention should be paid to a generation of more realistic human body models with detailed inner organ models in order to be able to describe the organ injury mechanisms during accidents. In this study the abdominal finite element (FE) model was created to improve the abdominal part of the human body model ROBBY. This should ensure the ability of an updated model ROBBY to predict the abdominal injury during various impact conditions. The abdominal model geometry was created on the basis of real anatomical human data. Tissue characteristics completing the organ geometry were obtained from public sources. The created abdominal model was consequently scaled and embedded into the ROBBY model representing a 50th percentile male. The updated ROBBY model was validated with a comparison of the simulation results during abdominal oblique impactor tests and abdominal frontal rigid bar tests with available cadaver experimental results. It was found that the updated ROBBY model with the new deformable abdominal model is able to predict the response of the human abdomen during various impact conditions very well. Moreover due to the precision of the abdominal model the injury of individual abdominal organs can be predicted during impact. This allows the simulation of real vehicle accidents and the assessment of human abdomen injuries arising during accident to be performed.

KEY WORDS: Biomechanical model, human abdomen, generation, validation, injury assessment.

1 INTRODUCTION

With development of computers and numerical methods during the last years, more complex virtual biomechanical models have been developed to study human body injury mechanisms in vehicular impacts (Haug et al., 2004; Iwamoto et al., 2002; Oshita et al., 2002; Ruan, 2005, and Zhao & Narwani, etc.). These models of various complexity show promise as useful tools for the better understanding of impact problems and therefore may reduce the dependence on cadaver experiments, which are expensive, time consuming, hard to repeat, etc. Moreover, the great advantage of a well validated deformable human model is the ability to predict the injuries to inner organs.

In the presented study a new abdominal FE deformable model was generated. The abdominal model was implemented into the human articulated rigid body model ROBBY (Hynčík, 1998), which was previously developed at University of West Bohemia in cooperation with ESI Group (Engineering Simulation for Industry).

ROBBY, which represents a model of the average adult man, is still being improved to obtain a more factual human body model with the possibility to predict organ injuries during vehicle accidents. This model belongs to the ROBBY family, which consists of ROBBY – a model of a 50th percentile male (on various levels (deformable, rigid)), ROBINA – model of a 5th percentile female and BOBBY – a model of a child. The first ROBBY model was created from rigid bodies only connected with joints (Hynčík, 1998). Consequently, muscles and some internal organs were integrated into ROBBY (Hynčík, 1998; Hynčík, 2002). Recently, a new FE deformable thoracic model based on real anatomical human data was created (Číhalová, 2006) and integrated into the ROBBY model (Číhalová & Hynčík, 2008). The aim of the presented study was to continue with the updating of ROBBY model to create a more realistic human body model with the possibility to predict organ injuries during accidents.

2 METHODS

The abdominal model generation (geometry creation, material and contact definition) and the investigation of its dynamical response during various impact conditions are described below.

2.1 The creation of abdominal organ geometry

The geometry arrangement of the new abdominal FE model was based on real anatomical human data. The photographs of cross-sections through the human body offered by VHP (Visible Human Project®, 2003) were used to construct the geometry arrangement of individual abdominal organs. These photographs were processed in advanced software for their visualization, Amira® (2002), which is able to create a 3D object model from the cross-sections through this object. In this software the boundaries of individual organs were marked in each photograph. Consequently, the tetrahedral meshes of individual abdominal organs were generated. The meshes were remeshed to the hexahedral ones in the Altair® HyperMesh® (2009) software, high-performance finite element pre- and post- processor for popular finite element solvers. The measurements of the individual elements representing the abdominal organs were chosen with respect to the stability of computation in the solver PAM − CRASH™ (2009).

The time step was held not to be smaller than 10^{-3} ms. The created abdominal organ models were scaled in accordance to anatomical reality (Gray, 1995). Consequently, the scaled abdominal model was integrated into the ROBBY model (Figure 1).

Figure 1: The thoracic and newly created abdominal segments of the ROBBY model.

Great attention was paid to the liver organ, which is a vital organ (production of bile, blood formation, manufacturing of anticlotting heparin, storage of proteins and sugar). This organ is very often injured during accidents. Since the liver is thickly interlaced by bloodvessels, the models of the inferior vena cava and portal vena were created on the basis of VHP photographs (Visible Human Project®, 2003) and anatomy (Gray, 1995). These were consequently integrated into the liver model (Figure 2).

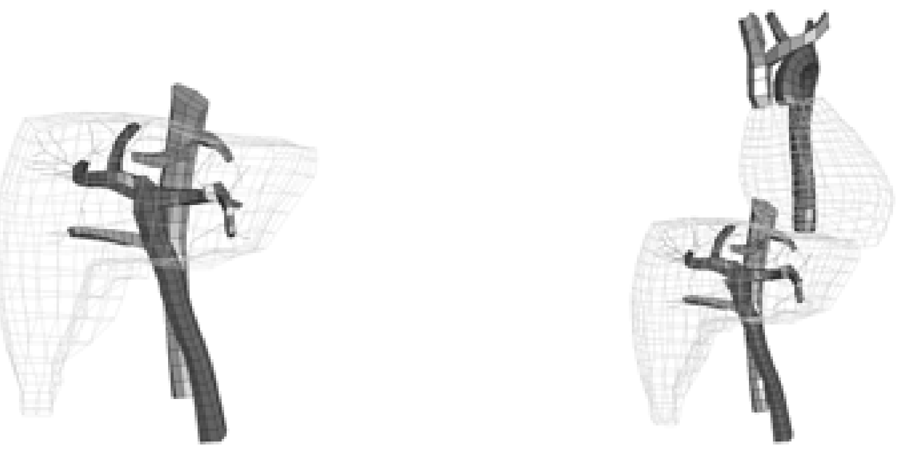

Figure 2: The blood vessels model inside the liver and its connection to the heart.

To define the correct linkage of blood vessels inside the liver, a grid of insignificant mass within the liver was generated. The vessel models were fixed by a tied contact offered by PAM−CRASH™ (2009) to this grid. Moreover, the vena cava was connected to the heart (Figure 2) according to anatomical reality (Gray, 1995).

2.2 Description of the abdominal model

The models of abdominal organs, such as the liver with the portal vena and the inferior vena cava, the spleen, the left and right kidney, intestines, and the stomach, were embedded into the new abdominal model. The urinary bladder was used from the previous ROBBY model. The thoracic and abdominal segments were separated by the diaphragm model. Some of the abdominal organs were located in the peritoneum model. The solid organs, such as the liver, spleen and kidneys, were modeled with a set of brick elements described by linear visco-elastic behavior (PAM-CRASH™, 2009). The hollow organs, such as the stomach and intestines, were simulated in two ways: i) by a set of brick elements described by linear visco-elastic behavior and ii) by a biobag (bag with the fluid inside) (Hynčík, 2002). One integration point scheme with viscous method prevention was used for solid elements (PAM-CRASH™, 2009). The reason for the use of the one point integration scheme was a four times lower time costliness in comparison to the eight integration point rule (PAM-CRASH™, 2009).

The biobag model derived from the airbag model (Haug et al., 2004) can be used in the FE modeling of organs. Its big advantage is the saving of computational time, since there are no finite elements describing the fluid inside. The fluid has only its state characteristics, which is constant in the whole volume surrounded by the biobag surface.

Constant atmospheric pressure and constant normal body temperature were applied to initialize the biobag. The isothermic process was supposed, since there is no considerable change of temperature in the human body. The material characteristics of the fluids inside the biobags were obtained in cooperation with the ESI Group.

All abdominal organs were covered by thin layers represented by the set of shell elements. One integration point scheme with stiffness method prevention (PAM-CRASH™, 2009) was used. The reason for the use of the one point integration scheme was a 2.5 times lower time costliness in comparison to the four integration scheme use.

The linear elastic isotropic material model offered by PAM−CRASH™ (2009) was used to capture the behavior of abdominal organ layers and of the peritoneum and diaphragm.

Two types of contacts offered by the solver PAM-CRASH™ were used to simulate the mutual contacts of all abdominal organs. The first type of used contact is the sliding contact (PAM-CRASH™, 2009). This contact allows two interfaces to slide with respect to each other. The second one is the tied contact (PAM-CRASH™, 2009), where one part is fixed at a certain distance (describing the reality) with respect to a second one. This contact is used to simulate ligaments or firm connections among organs. Particular examples of such contact in the abdominal model are the simulation of the close conjunction of the liver to the diaphragm or for fixation of blood vessels inside the liver.

2.3 Validation of the abdominal model

The updated model ROBBY, i.e., the ROBBY model with the new abdominal model, was validated against a pendulum impactor test according to Viano (1989) and Viano et al. (1989) (Figure 3) and a rigid bar test according to Cavanaugh et al. (1986) (Figure 3). These two tests approximate the most common situations during a traffic accident in laboratory conditions. The first one approximates the oblique impact, when the upper abdomen comes to contact with side door. The second one approximates the frontal impact, when the lower abdomen comes into contact with the bottom part of the steering wheel. The validation results of simulations were compared with cadaver experimental data according to Cavanaugh et al. (1986), Viano (1989) and Viano et al. (1989). In both test types, the solid and biobag variants to model hollow abdominal organs were used.

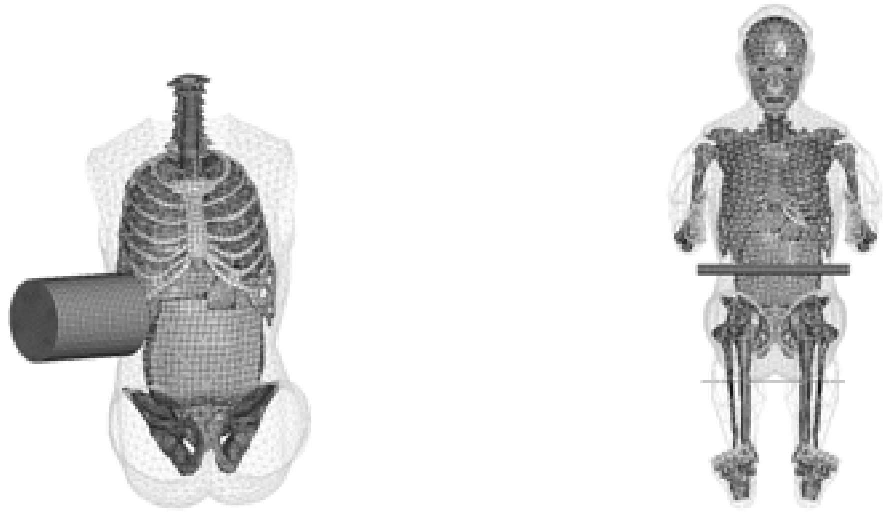

Figure 3: Initial position of the ROBBY model in the pendulum impactor test according to Viano and in the rigid bar test according to Cavanaugh.

2.3.1 Model validation against Viano test

Data of cadaver studies presented by Viano (1989) and Viano et al. (1989) were used to validate the upper part of the abdominal model in the oblique direction.

Two tests with various initial conditions: i) Impactor of mass 23.4 kg impacting the abdomen at speed 4.79 m s^{-1} and ii) Impactor of mass 23.4 kg impacting the abdomen at speed 8.32 m s^{-1}, were performed.

The force-deflection dependencies were investigated and these were compared with the experimental results (Viano, 1989; Viano et al., 1989). The force value was taken as the impactor force (Haug et al., 2004) in the impact direction (Viano, 1989). The force was filtered by Channel Frequency Class (CFC) 600 (PAM-CRASH™, 2009). The abdominal deflection was considered to be the difference of total displacement of the impactor and thoracic vertebrae T12 (Viano, 1989), i.e., displacements in all three directions were taken into account.

2.3.2 Model validation against Cavanaugh test

Data of cadaver studies presented in Cavanaugh et al. (1986) were used to validate the lower part of abdominal model in the anterior-posterior direction.

Two impacts were performed with various initial conditions: i) low energy impact (Impactor of mass 32 kg impacting the abdomen at speed 6.1 m s^{-1}.) and ii) high energy impact (Impactor of mass 64 kg impacting the abdomen at speed 10.4 m s^{-1}.).

The force-deflection dependencies were investigated. The abdominal deflection was taken as the difference of horizontal displacement between the impactor and L3 (Cavanaugh et al., 1986). The force value was considered to be the impactor force in the impact direction (Haug et al., 2004). The force was filtered by CFC 600.

3 VALIDATION RESULTS

3.1 Validation results of Viano test

Figure 3 compares the simulation results of force-deflection dependencies of the abdominal model for both organ model variants with experimental results during right side impacts.

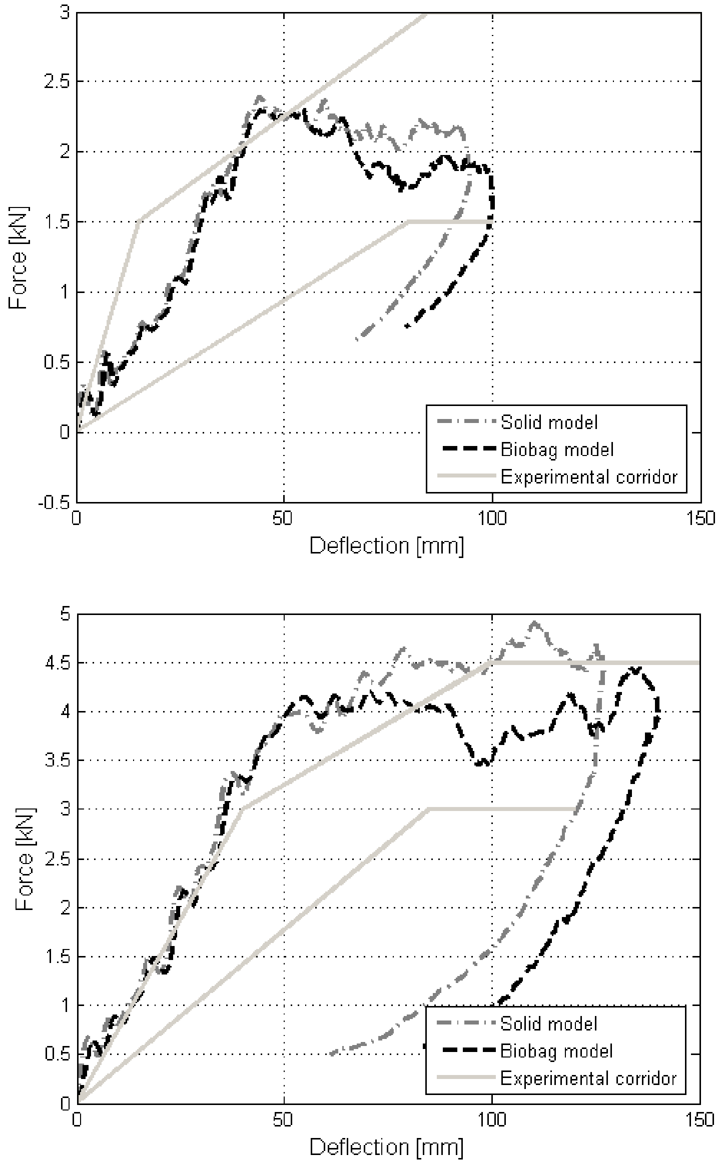

Figure 3: The validation results during the oblique impactor test at 4.79 m s^{-1} and at 8.32 m s^{-1}.

The force-deflection dependencies, which are hysteresis curves, are characterized by their loading, plateau and unloading part. The loading part is characterized by a rapid rise, which is mainly caused by the viscous behavior of the upper abdomen. The plateau region is also the result of the viscous response. The unloading part represents the unloading of the compressed tissues and the elastic non-linear unloading of the upper abdomen follows. The shape of these three phases depends on the impact energy.

For the impactor pendulum test at impact speed 4.79 m s^{-1} the simulation results for the solid and biobag model are comparable with the experimental results, i.e., the results of the simulations fit the experimental corridor. The impacted force predicted by the abdominal model reaches its maximum during the lower deflection in comparison with the experimental results (Figure 3). This shows the higher stiffness of the upper abdomen model in comparison with a real human abdomen. The stiffness could be reduced by selecting different organ material parameters. However, since the simulation results are comparable with experimental results and due to the lack of time, this can be the subject for future studies.

For the impactor pendulum test at the impact speed 8.32 m s^{-1} the simulation results for the solid and biobag model are comparable with the experimental ones in all three phases: phase of loading, plateau and unloading. The impactor force reaches its maximum during lower deflection in comparison with the experimental results, as well as during the 4.79 m s^{-1} Viano test (Figure 3).

3.2 Validation results of Cavanaugh test

The simulation and the experimental results of the low energy and high energy Cavanaugh test can be seen in Figure 4. The loading part of the experimental force-deflection curve of the lower abdomen of the human cadaver can be characterized by an almost linear rise from zero to peak force. The unloading part of the experimental force-deflection curve of the lower abdomen is the approximately vertical line (Cavanaugh et al., 1986).

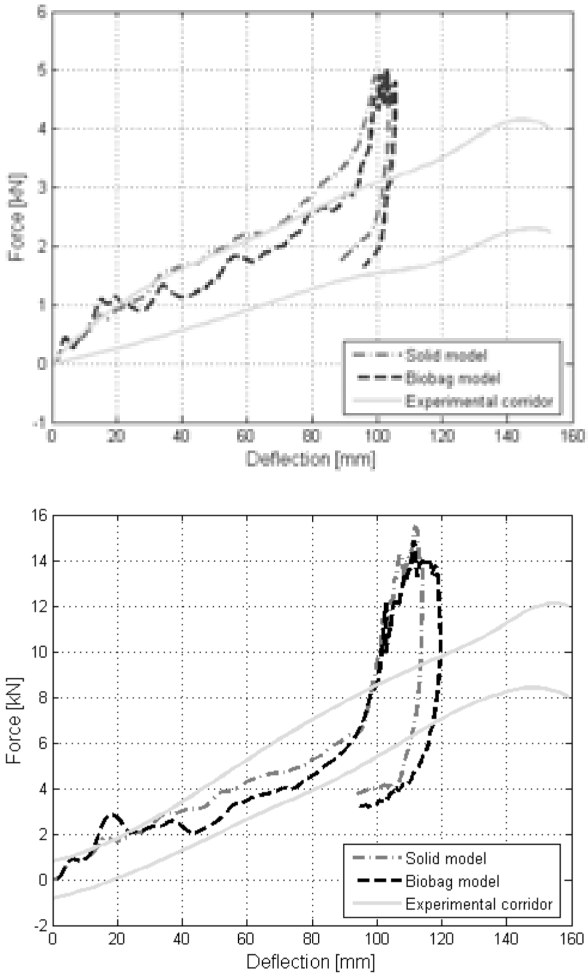

Figure 4: Results of validation during the frontal rigid bar test at 6.1 m s^{-1} and at 10.4 m s^{-1}.

In both rigid bar tests the results of the simulation are comparable with the experimental ones. In both cases there is a problem around deflection of approximately 90 mm, when the force increased rapidly.

In summary, the comparison of the simulation and experimental results for the lower and upper part of the abdomen shows that the created abdominal model is able to approximate the abdomen response of the cadaver sufficiently.

4 DISCUSSION

Computer numerical based simulations are presently frequently used to analyze the crash situation of a vehicle in detail. On the other hand the detailed analysis of passenger injuries is limited. This is caused by the fact that the numerical dummy models do not permit the study of inner organ injuries. The possibility arises of using FE human body models with integrated inner organ models. However, such human body models contend with the problem of obtaining parameters describing organ behavior and arising injuries.

Our department has been developing the FE human body model ROBBY for a long time. This model is still being improved in order to be able to describe various crash situations. In this study the new deformable model of abdomen was developed. The geometry of model was based on data offered by the VHP (Visible Human Project®, 2003). Material parameters completing the inner organ models were obtained from Ruan (2005); Ruan (2003) and Zhao & Narwani. The division of the abdominal organs into two groups: i) hollow organs and ii) solid organs was taken into account. To capture this reality, the modeling by a set of solid elements or by a biobag (sac with fluid characterized by state relation inside) was applied. The abdominal model was validated by a frontal rigid bar test at the lower abdomen part according to Cavanaugh et al. (1986) and by an oblique impactor test at the upper abdomen part according to Viano (1989) and Viano et al. (1989). Figures 3 and 4 summarize the results of the validation. It can be concluded that the newly created abdominal model was validated satisfactorily, i.e., the results of the simulations correspond to the experimental ones.

In the case of the Cavanaugh tests (Figure 4) an abrupt increase of force in the deflection of approximately 90 mm can be observed. This increase is at the moment when the compressed abdominal cavity starts to push on the lumbar spine. Hence the cause of the sudden growth is the greater stiffness of the lumbar spine. This problem was beyond the scope of this study. However, a new model of the lumbar spine, based on the real anatomical human data offered by VHP, will be created in the near future. This model will be completed by suitable material data and consequently it will be validated. It can be noted that the influence of the spine is not apparent in the case of the Viano test (Figure 3). This is caused by the fact that the abdomen deformation is smaller in the case of Viano test than in the case of the Cavanaugh test. This corresponds with the reality that there is no contact of the impactor with the ribcage in the case of the Cavanaugh test in comparison with the Viano test, i.e., a bigger deflection of the abdomen arises in comparison to the Viano test.

5 CONCLUSION

Dynamical abdominal responses during the frontal rigid bar test and the oblique pendulum test were investigated using the human body model ROBBY with a newly created abdominal model. The model was validated by a comparison of force-deflection dependencies obtained from simulations with experimental results. This validation is satisfactory. The abdominal model can be used to predict organ injuries during various accidents.

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