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Evaluation of the effect of muscle forces implementation on the behavior of a dummy during a head-on collision

KAMIL SYBILSKI[1], ŁUKASZ MAZURKIEWICZ[1], JACEK JURKOJĆ[2], ROBERT MICHNIK[2], JERZY MAŁACHOWSKI[1]

[1] Institute of Mechanics and Computational Engineering, Faculty of Mechanical Engineering, Military University of Technology, Warsaw, Poland.
[2] Department of Biomechatronics, Faculty of Biomedical Engineering, Silesian University of Technology, Zabrze, Poland.

Purpose: The aim of this study was to develop a method to implement muscle forces to a numerical model of a dummy and to evaluate the effect of muscle activation on driver behavior during a frontal collision. The authors focused on the forces acting at the knee, hip, and elbow joints. *Methods*: The authors carried out torque measurements in joints using the Biodex System 4. Then, the previously developed numerical models were modified by introducing the joint torque values. Moments of force were introduced as a function of the rotation angle. During research, numerical simulations were carried out in three stages: in the first stage, a full vehicle crash was analyzed to determine the change of velocity of the vehicle interior; in the second stage, subsidence of the system was realized; in the third stage, a frontal crash was simulated. The models considered the operation of the sensors, airbag and seat belt tensioning system. *Results:* A numerical model with the active response of the dummy to the change in position during impact was developed. The results of the dynamic analysis were used to analyze the impact of muscle activation on dummy behavior. The change in shoulders rotation angle, the lateral and vertical displacement of the dummy's center of gravity, and the forces acting between the dummy and the seat belt were compared. *Conclusions:* The effect of muscle action on the behavior of a dummy during a frontal collision was determined.

Key words: *muscle forces, dummy, driver, finite element method, safety, biomechanics*

1. Introduction

The development of numerical methods has made numerical analysis one of the main research methods in the human safety assessment process. One of the main areas of application of numerical methods in this field is all kinds of simulations of dangerous situations on the road [15]. Researchers' works are focused on practically every aspect and object that can influence on increasing the accuracy of the obtained results or, on the contrary, on decreasing the time of calculations, very important in case of, for example, optimization analyses. Some of the works are focused on the construction of numerical models of vehicles [24], others on the construction of numerical models of infrastructure being the environment of the road event [33]. Not meaningless is also the development of the commercial models of dummy family [7], [13], [31], thanks to which it is possible to analyze the injuries of any road traffic participant, including children [2], [20]. Dummy models are also used to assess injuries resulting from factors other than dangerous driving situations. These include all types of injuries sustained during armed conflicts [9], [27].

The use of very accurate dummy models significantly increases the computation time, requires high computational power, and, in many cases, makes it impossible to perform the calculations. Therefore, it is quite common to perform numerical analyses using simpler models to which an accurate model of, for example, a selected organ is attached and appropriate loads are transferred to it. This gives the possibility of a very accurate tracing of injuries of the selected body part but requires prior development of its very accurate numerical model. Examples are works focusing on mapping real injuries of the head [5], [26], [34], pelvis [16], spine, and chest [14], [23].

* Corresponding author: Kamil Sybilski, Institute of Mechanics and Computational Engineering, Faculty of Mechanical Engineering, Military University of Technology, ul. gen. Sylwestra Kaliskiego 2, 00-908, Warsaw, Poland. Phone: +48 261 83 96 83, e-mail: kamil.sybilski@wat.edu.pl
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The most advanced dummy and human body models for crash analysis are currently GHBMC (Global Human Body Models Consortium) [35] and THUMS (Total HUman Model for Safety) [8]. These are numerical models that fully reflect internal organs and muscle action. Modeling of internal organs allows for a complete analysis of injuries during an accident. On the other hand, modeling of muscles and their actions during the accident allows to determine how the human body will behave in reaction to external factors [6], [22]. Unfortunately, the high complexity of the models goes together with the high demand for computing power, which results in very long computation times. Moreover, even such complex numerical models are not able to perfectly reproduce the behavior of the human body. Among others, it has been shown in [21] that in THUMS v3 or GHBMC models, the fat material models of fat are not sufficiently accurate to describe its real behavior during the interaction between the pelvis and the waist belt. On the other hand, the authors of the paper [14] point to overestimation of the rib cross-sectional height by 22% (Thums) and 18% (GHBMC M50-O) which affects the overestimation of its bending stiffness. Therefore, it is important to continue work on the development of both accurate numerical models of the human body, as well as numerically efficient models, i.e., considering the key aspects in a given analysis and, at the same time, allowing to analyse many variants in a short time. Numerical models of dummies used, for example, in crash-tests enabled allow us to perform much less time-consuming dynamic analyses and to determine basic biomechanical parameters that enable to estimate injuries during crashes. However, the main disadvantage of these models is the lack of representation of active and not only passive, passenger response to overloading. The response in the form of muscle tension, for example, of the neck or upper and lower limbs, can significantly affect human behavior during a crash [6]. In this paper, the authors attempted to implement muscle forces into a numerical model of a Dummy Hybrid III 50th Humanetics dummy [13].

2. Materials and models

2.1. Biomechanical measurements

The purpose of this study was to develop a method for implementing muscle forces in a numerical model of the Dummy Hybrid III 50th dummy [13], which is commonly used for analysis of automotive crashes. The authors decided to use the measured characteristics of the moments of force in the joints as a function of the limb's rotation.

During the experimental study, the authors focused on the elbow, knee, and shoulder joints. A Biodex System 4 device was used to perform the torque measurements (Fig. 1). Measurements were performed for three men of age 24–30 years, weight 76–78 kg, height 172–177 cm. The first two subjects were physically active (sport 2–3 times a week). The third person did not do any sports and had a sedentary lifestyle. All participants performed a five-minute warm-up before data acquisition. Measurement protocols for each joint include the determination of torque in five angular positions, corresponding to the anatomical rotation contours in the respective joint. For the elbow joint, the sample angles were 0, 30, 45, 60 and 90°; for the knee joint: 15, 30, 45, 60 and 90°; for shoulder flexion/extension: 60, 75, 90, 105, and 120°, and adduction/abduction: 0, 5, 15, 30, 45°. These values were chosen to coincide with the range of angles occurring in the dummy limbs during impact.

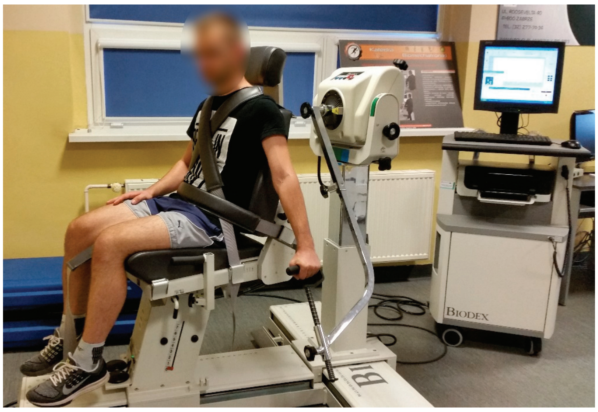

Fig. 1. Measurement of torque in the shoulder joint

2.2. Numerical analyses of frontal crash

In the second phase of the study, numerical analyses of a head-on collision of a car with an initial speed of 50 km/h were performed. At this stage, the authors used the previously developed methodology to prepare numerical models and conduct simulations studies [28], [29]. During numerical analyses, the authors analyzed the case of a driver with reduced mobility of the left hand and significant paralysis of the right hand, acquired as a result of tetraplegia. With this disability, the driver must use a special three-bolt steering wheel grip. Some of the driving tasks, such as switching on the lights or turning signals, can be carried out via buttons on the armrest, for example.

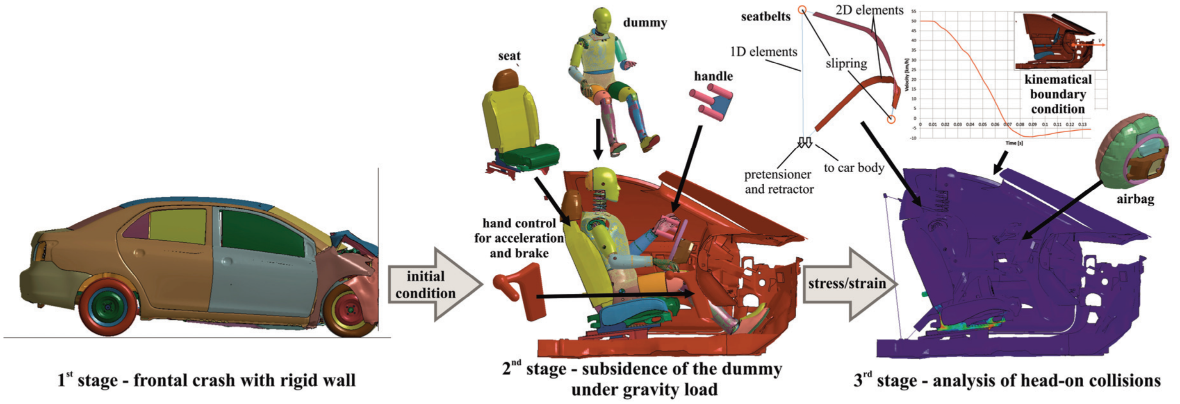

Fig. 2. Numerical research strategy [29]

The aforementioned simulations were conducted in three stages (Fig. 2). The first includes a full-frontal collision of the vehicle with a rigid wall. The results of this stage, the velocity of the vehicle interior versus time, are used as boundary conditions in the third stage. The second stage involves the subsidence of the dummy and the typical interaction with the vehicle interior components. The dummy was placed close to the target position, a few millimeters above the floor, seat, and additional instrumentation. Subsequently, a dynamic analysis with gravity loading was performed. The analysis considers the full interaction of the dummy with the interior of the vehicle, including the occurrence of friction and deformation of structures.

In the third stage of analysis, the numerical model from the second stage was used in deformed state with residual stresses included. This model was completed with a seatbelt system with a functioning retractor and pretensioner, an airbag and sensors to activate the operation of the airbag and belts. The initial-boundary conditions were also introduced, an initial velocity of 50 km/h was applied to the entire model and the velocity driven motion of the vehicle body was used based on the results obtained in the first stage.

In the third stage, two independent numerical models differing in the implementation of muscle force constitutive models were analyzed. The first model was the original model provided by Humanetics [13], without any muscle forces defined. In the second model, the CONSTRAINED-JOINT definition [10], [17], responsible for the rotation of the different parts of the dummy's limbs, was modified. The modification was to introduce the joint torque defined as a function of the rotation angle. The corresponding curves were prepared based on experimental measurements according to the scheme shown in Fig. 3. For each joint, a different curve with independent values for flexion and extension was prepared. The angle of the neutral joint was defined based on the desired relative positions of the limbs. In the case of forcing a small rotation, a proportionally increasing torque was generated in the joint up to the average value obtained from the biomechanical measurements. In the case of larger rotations, a moment with a value corresponding to the average of the measurements acted. Thus, movement was possible when external loads caused a load greater than that defined in the joint.

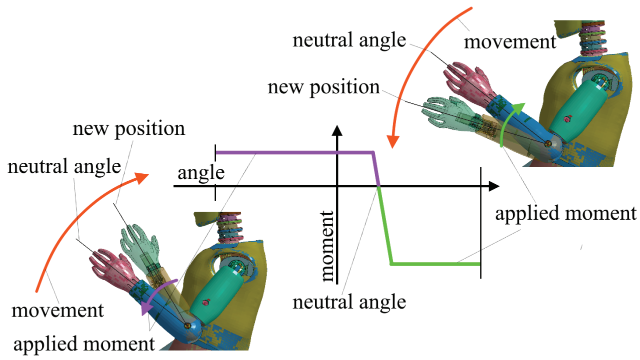

Fig. 3. Implementation of force moment characteristics in joints

In both models, the position of the dummy at the beginning of the analysis was the same. All numerical analyses for the numerical simulation of the dynamic cases using the LS-Dyna solver with an explicit integration step to solve the dynamic equilibrium equation were performed.

3. Results

3.1. Measurement of moments of force in joints

The purpose of the first step was to determine the torque generated at the selected joints for the three subjects. The results of the measurements are presented in Tables 1–4.

Table 1. Results for the elbow joint [Nm]

Side		Right			Left		
Person		1	2	3	1	2	3
Extension	0°	42.9	14.1	28.4	49.6	46.5	24.6
	30°	60.3	56.6	38.0	51.6	53.1	39.0
	45°	64.4	61.5	36.3	60.8	48.6	39.9
	60°	64.6	55.0	38.3	65.2	42.7	41.2
	90°	56.7	45.8	36.7	58.9	33.6	39.0
Flexion	0°	68.8	53.9	54.4	49.5	52.0	48.4
	30°	54.2	45.2	56.4	50.8	64.6	54.6
	45°	59.3	53.9	60.1	61.4	67.1	56.0
	60°	57.1	63.5	59.7	65.1	63.0	56.7
	90°	52.4	64.3	60.2	61.9	51.2	53.2

Table 2. Results for knee joint [Nm]

Side		Right			Left		
Person		1	2	3	1	2	3
Extension	15°	89.8	122.3	67.9	136.1	107.8	76.7
	30°	145.4	150.9	112.0	168.7	20.9	98.1
	45°	180.2	211.8	146.7	180.2	168.7	120.4
	60°	216.0	267.4	174.2	270.3	249.3	145.9
	90°	310.8	266.7	178.0	250.2	259.6	178.8
Flexion	15°	116.5	90.1	53.2	130.5	80.7	72.0
	30°	111.5	107.2	43.6	117.7	96.3	73.0
	45°	102.6	122.7	49.4	100.8	115.6	57.1
	60°	87.6	118.1	56.3	97.0	104.9	93.4
	90°	82.3	87.9	65.6	84.9	70.8	69.5

Table 3. Results for the shoulder joint [Nm]

Side		Right			Left		
Person		1	2	3	1	2	3
Abduction	0°	65.1	88.5	58.5	68.4	66.3	63.1
	5°	62.4	87.9	62.4	62.2	63.7	68.6
	15°	70.3	69.1	58.7	59.7	56.8	53.8
	30°	74.7	67.5	64.3	69.3	59.8	67.1
	45°	72.4	56.9	62.7	69.6	58.1	59.2
Adduction	0°	67.8	77.6	48.0	52.2	44.1	60.1
	5°	67.1	78.2	49.6	50.1	54.3	58.1
	15°	77.0	59.6	60.6	63.6	63.1	65.6
	30°	76.5	75.0	60.6	73.3	64.8	71.0
	45°	73.2	82.1	67.9	70.3	72.6	74.1

Table 4. Results for the shoulder joint [Nm]

Side		Right			Left		
Person		1	2	3	1	2	3
Extension	60°	56.0	64.3	35.3	54.8	57.1	52.1
	75°	57.9	62.1	37.4	59.1	50.4	44.0
	90°	57.9	59.2	43.1	60.1	51.0	40.7
	105°	57.8	58.6	48.4	61.2	47.1	46.1
	120°	59.0	54.0	57.9	63.0	50.4	47.7
Flexion	60°	59.8	72.9	48.6	60.2	56.2	56.0
	75°	62.4	74.1	52.2	59.7	56.9	54.9
	90°	67.2	76.6	56.0	58.6	61.7	56.8
	105°	67.2	70.8	54.1	52.1	60.3	54.2
	120°	68.0	57.2	54.7	53.8	55.0	52.8

The averages were calculated for each joint angle value and are shown in Figs. 4–7.

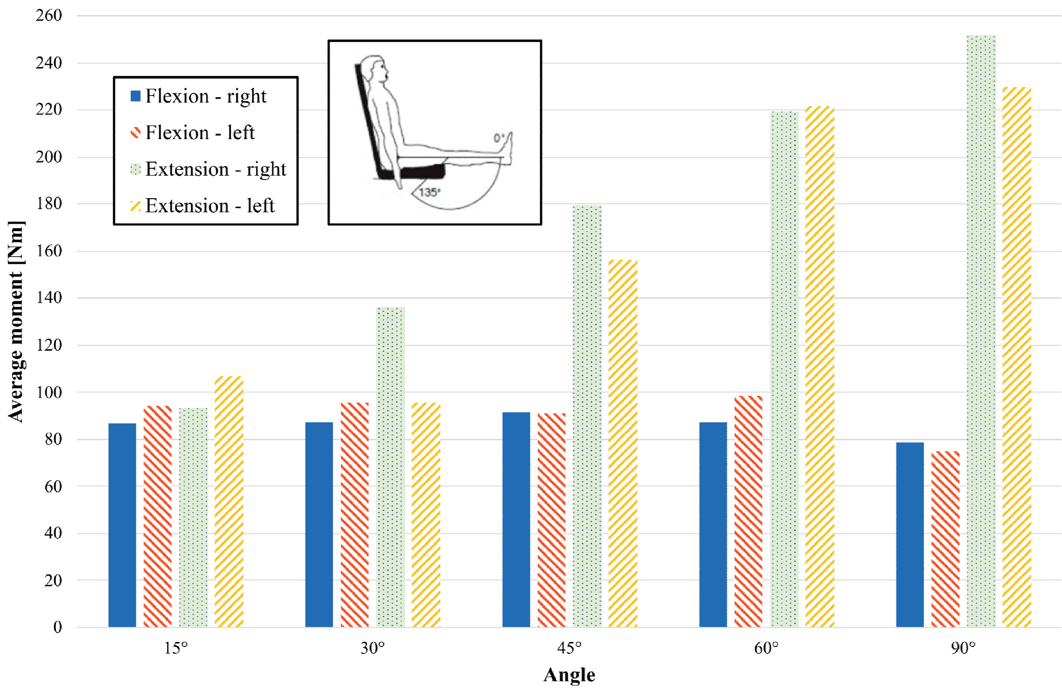

Fig. 4. Average knee joint force moment

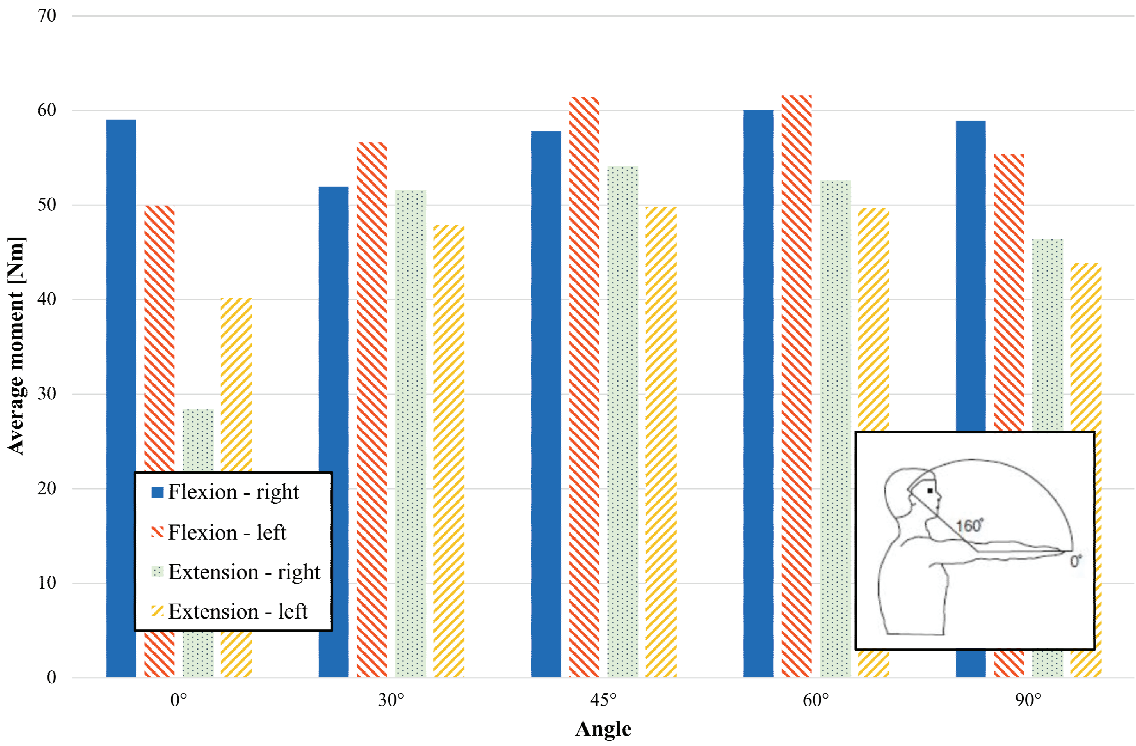

Fig. 5. Average elbow joint force moment

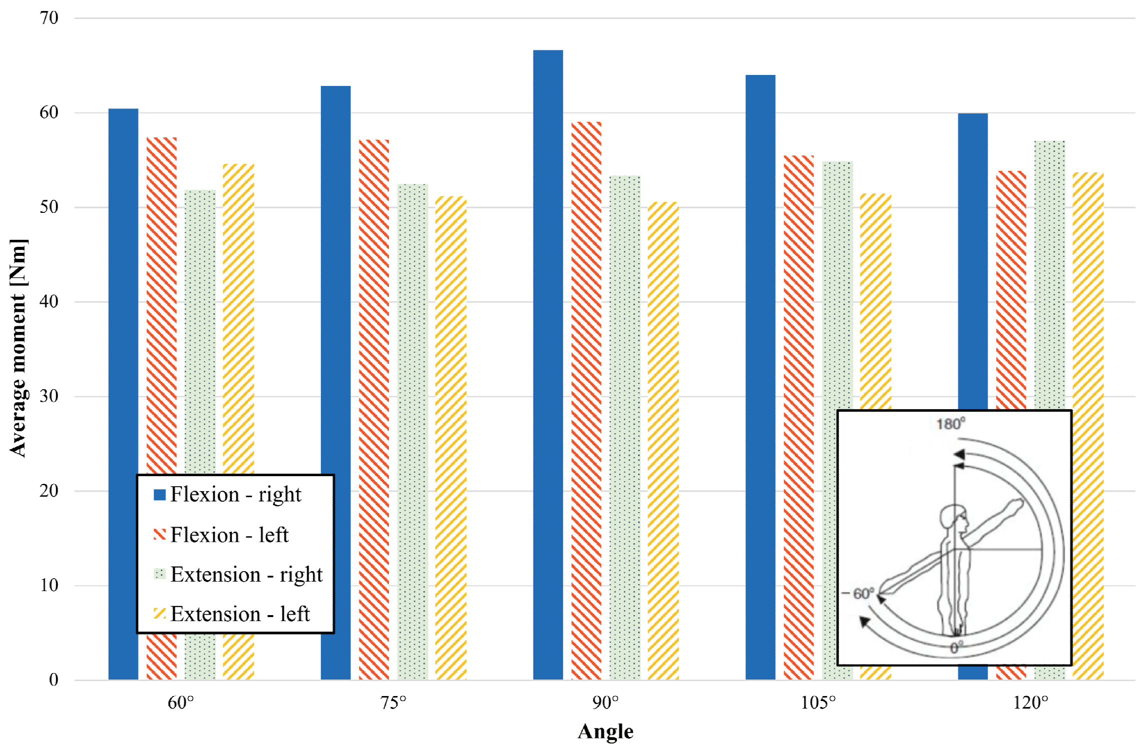

Fig. 6. Average shoulder joint force moment

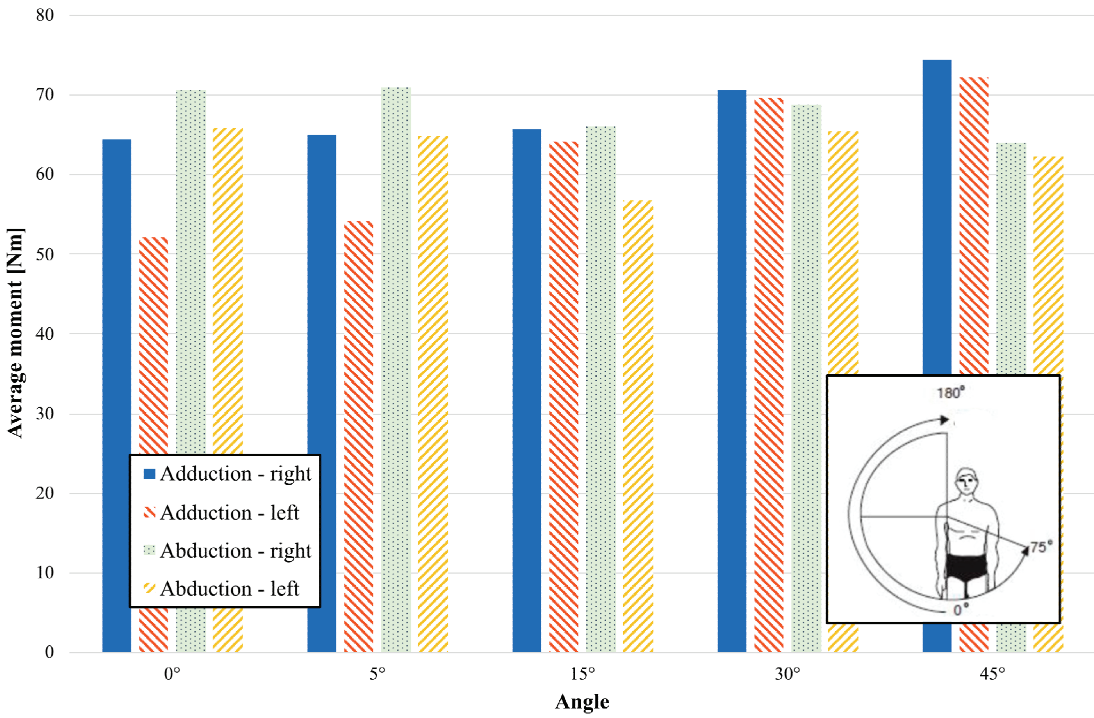

Fig. 7. Average shoulder joint force moment

3.2. Numerical analysis

The course of analysis in the first and second stage was described in the earlier authors' works [28], [29]. In the third stage, for both models, the course of simulations was very similar to each other. The subsidence analyses were completed at time $t = 0.3$ s, and this is also the starting point of the analysis in the third stage. After about 10 ms, the sensors activate the airbag and the seatbelt tensioning system. At time instant $t = 0.316$ s, the opening of the airbag cover is visible. At time instant $t = 0.336$ s, the airbag is fully inflated, causing the hand together with the handle on the steering wheel to be pushed outwards. After 57 ms from the beginning of the analysis, the dummy's head impacts the airbag and is then pushed increasingly into the airbag over the next 20 ms, reaching maximum

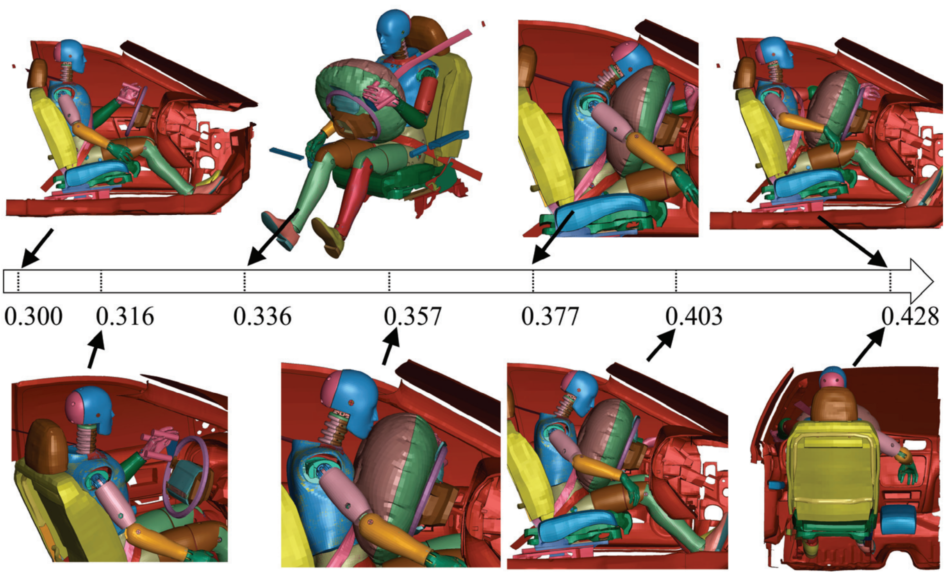

Fig. 8. Frontal crash stages

forward tilt at t = 0.377 s. From this point on, head movement in the opposite direction begins (the pelvis reaches maximum longitudinal displacement at time instant t = 0.367 s, 10 ms earlier). At time t = 0.403 s, the head stops contacting the airbag, and at time instant t = 0.428 s, the head hits the headrest. The introduction of muscle tension does not have a significant effect on this process; the differences in the occurrence of the stages between the two cases are less than 1 ms.

The introduction of muscle forces, on the other hand, influences the interaction forces that occur between the dummy and the seatbelts, which are shown in Fig. 9. In the case with muscle forces, the maximum contact force between the dummy and the seatbelts was approximately 4.2% less. The graph also shows that this force occurs at a slightly lower time.

In Figure 10, the change in shoulder rotation angle for both analyzed cases is shown. During the numerical analysis, the authors analyzed the case of a disabled

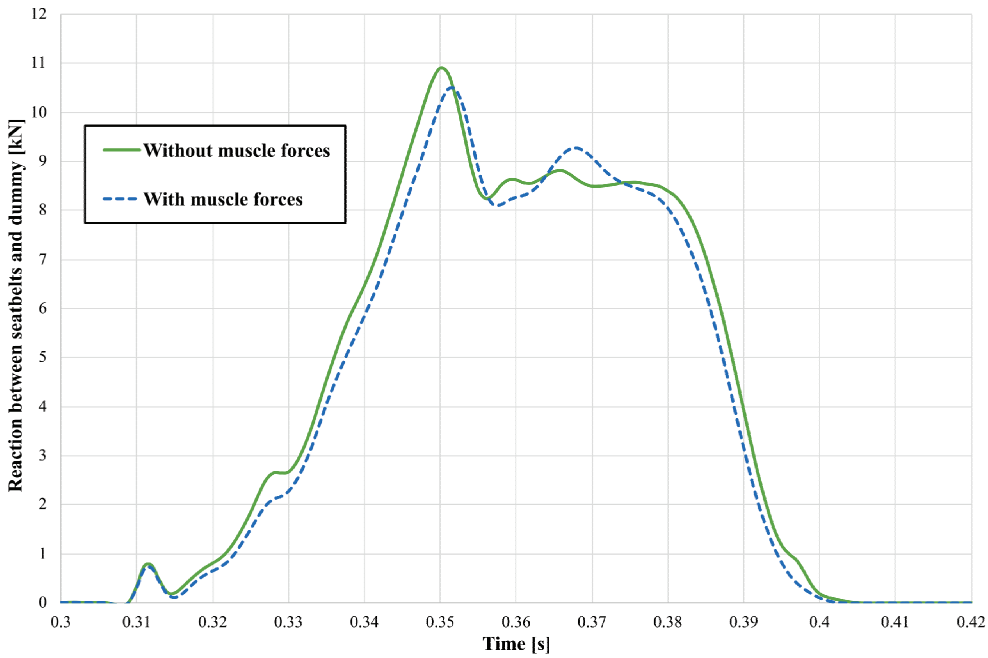

Fig. 9. Forces between seatbelts and dummy

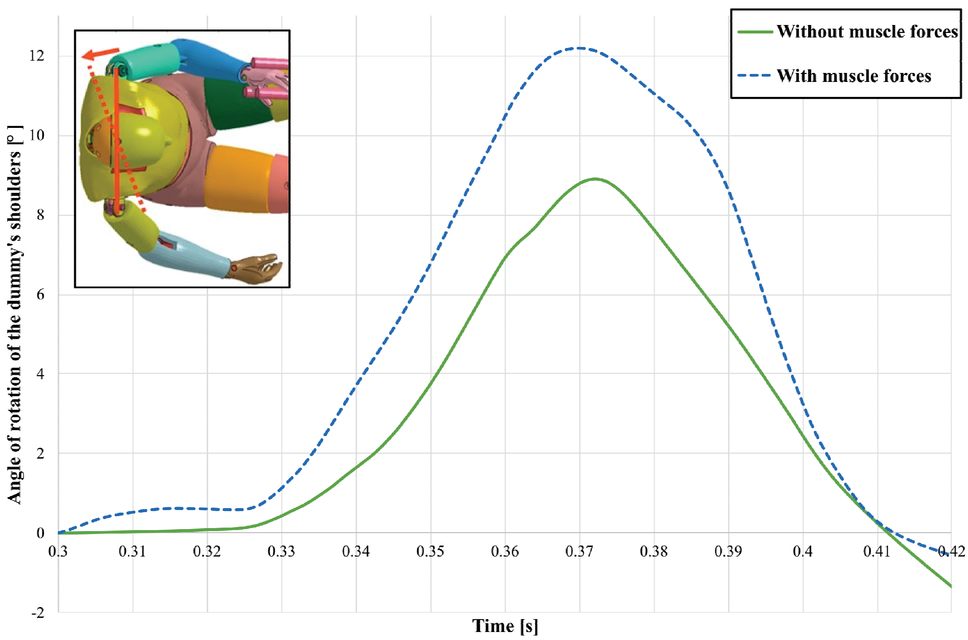

Fig. 10. Angle of rotation of the dummy's shoulders

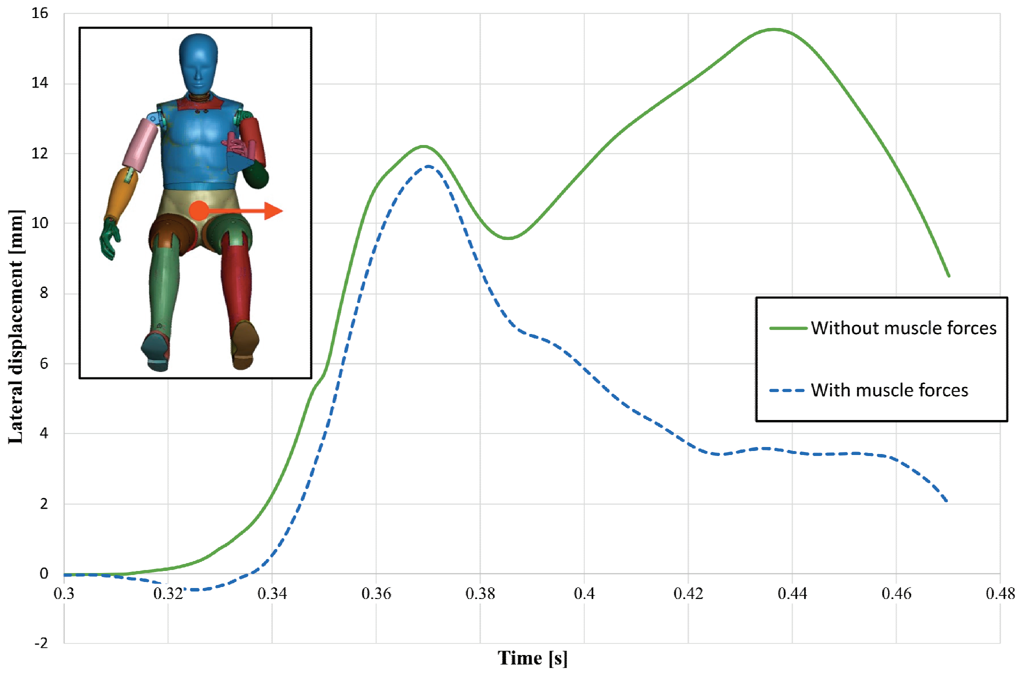

Fig. 11. Lateral displacement of the dummy's pelvis

driver who uses a special handle on the steering wheel for his left hand, with his right hand freely placed on the armrest (Fig. 8). This is a very asymmetrical case, and it is exacerbated by the presence of an asymmetrical three-point seatbelt. The additional introduction of muscle forces in the left arm increased this asymmetry and in the analyses, the shoulder rotation angle was significantly greater in the case with the muscle forces applied.

In Figure 11, the lateral displacement of the dummy pelvis is shown. Up to the moment of maximum longitudinal displacement of the head (t = 0.377 s), the shift of the curve for the case with acting torque in the joints is visible (the displacement occurs later and is slightly smaller). On the other hand, from time t = 0.377 s, it can be observed that the displacement of the dummy's center of gravity for the case without torque in the joints increases and for the case with

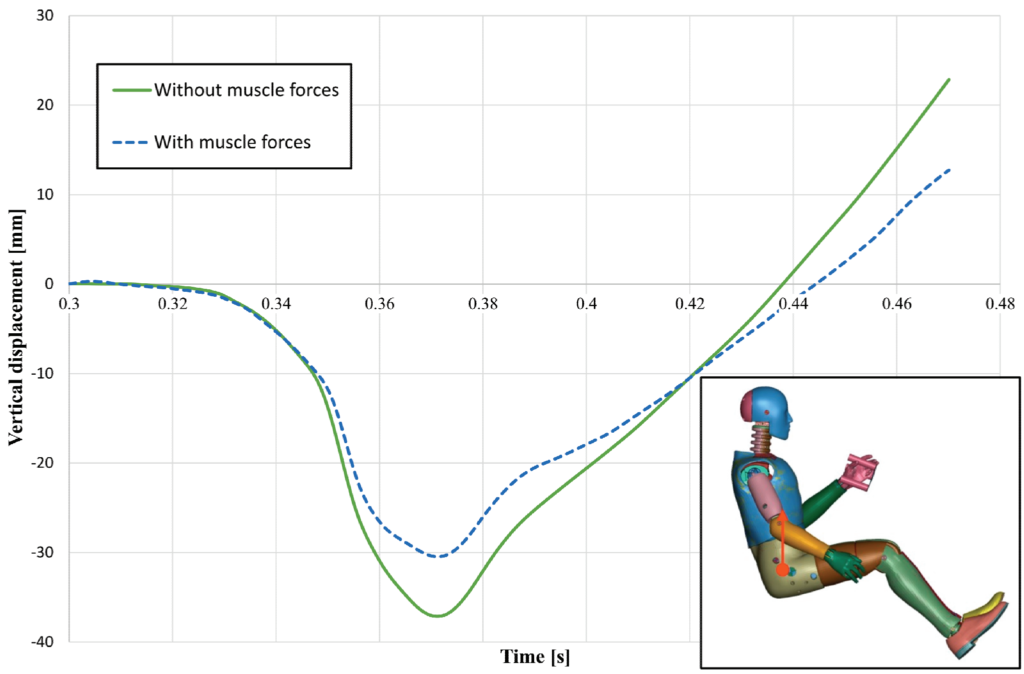

Fig. 12. Vertical displacement of the dummy's pelvis

implemented muscle forces decreases. The maximum difference is grater than 12 mm and is reached at time $t = 0.438$ s.

The same is observed for vertical displacement of the pelvis (Fig. 12). For the variant with implemented muscle forces, the vertical displacements are smaller.

4. Discussion

The assessment of injuries sustained during dangerous driving situations, including crashes, is now the subject of many studies worldwide. Undertaking this topic by many independent research groups contributes to a rapid increase in general and detailed knowledge of almost every aspect related to road user injuries.

Much of the ongoing research is concerned with driver posture and driving position. The authors of [18] noted that statistically women are smaller than men and are more likely to choose smaller cars, which, in turn, are more likely to cause injury during an accident. Thus, from the available statistical data, it is not possible to clearly determine to what extent posture affects the injuries sustained. Numerical analyses, which can analyze the influence of a selected parameter on the injuries sustained, in this case are a very helpful tool. However, this requires appropriate modeling methods and adaptation of the available models to the required conditions. In the field of posture influence assessment, many authors focus on methods of adjusting dummy models [25], [36] (mostly by morphing). The authors of works [4], [25], [36] have analyzed, among others, the influence of different heights and weights on the loads and injuries that occur in the body. The work [30] focused on different seat designs and numerical models and different positioning of the dummies. In addition, different seatbelts arrangements were analyzed. The result is an assessment of lumbar spine injuries. The works [1], [32] analyzed the effect of seatbelts positioning and its characteristics on the behavior of the dummy. However, in all the works described above, the displacement of the dummy and its rotation were not analyzed, and the analyzed crash cases focused on cases with both hands on the steering wheel and three-point seatbelts.

The asymmetrical positioning of the dummy during the crash was analyzed in the work [3], [11], [12], [19]. The authors reflected the asymmetry of body alignment by introducing the changes that occur in body positioning during road maneuvers and then determined its effect on the injuries sustained. Furthermore, the work of [3] considered muscle action using 1D elements and Hill material models. The center of gravity displacement and trunk were not analyzed in this work.

Despite the wide range of studies described in the literature, there is no clear reference to the results described in this paper. A reference can be the paper [29], which analyzed the crash of a disabled driver using a specialized handlebar mount and four-point symmetrical seatbelts with different characteristics. That paper used the same method to conduct numerical analyses and the same numerical models of safety systems.

The paper described a case with three-point seatbelts that introduced asymmetry in the support of the driver's body by blocking only the left shoulder. Additionally, the driver had his left hand placed in a grip on the steering wheel, giving an additional point of support to the left side of the body. The other hand was freely resting on the armrest without causing any resistance to upper body rotation. When comparing the results obtained with the results published in [29], it can be stated that resigning from the use of the four-point seatbelts causes almost twice the rotation of the shoulder. Analyzing the results obtained, it can be stated that the inclusion of muscle forces additionally increases this rotation by about 34%. The introduction of muscle forces also causes a significant change in the lateral displacement of the center of gravity.

The aim of this study was to develop a method for implementing muscle forces in numerical models of a dummy. In the literature on this topic, one can find a great number of works, mainly focusing on full 3D muscle models or on simplified 1D elements [22]. In both cases, it is necessary to adequately prepare the data allowing to change the muscle activation level, which is not a trivial task. In the proposed method, the data obtained from the measurements of the maximum moments of force in the joints are introduced directly into the numerical description of the joint, which is a significant simplification and acceleration of the work.

5. Conclusions

The aim of this study was to perform measurements of force moments at selected joints and to develop a method for their implementation into a numerical model of a dummy, which is commonly used for head-on collision analysis (the approach is based on a commercial Dummy Hybrid III 50th dummy model).

In the first stage, the moments of muscle forces acting at the selected joints for three persons were determined. The values obtained for each person vary quite a lot. It is influenced by the physical activity of the person and the dominant limb (two right-handed and one left-handed people were tested). In the second stage of the research, numerical simulations for the case without and including the action of muscle forces were performed, and based on this, the kinematic parameters of the dummy were determined.

Based on the presented results, it can be concluded that the influence of muscle action affects the behavior of the dummy, but not to a significant extent. One reason is the preliminary nature of the work and the limitations of the number of joints torques implemented. The work could be developed to include muscle activation in a larger number of joints. In addition, a great addition to biomechanical studies would be the ability to perform measurements under conditions of strong stimulation of the nervous system, comparable to the experience of a real accident. This would allow to obtain larger, more realistic values of the forces acting in the joints.

The approach described in this paper is a very good alternative to multibody analyses, which do not consider the deformability of the dummy. It allows for quick analysis of many variants of the impact while preserving deformability of the dummy structure and considering different variants of muscle action. The developed model is also a good alternative to advanced dummy models, e.g., Total Human Model for Safety (THUMS), which considers the muscle action, but the calculation time of a single case exceeds 5 days.

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