ON MODELING OF PEDESTRIAN IMPACT

Luděk Hynčík*, Hana Kocková*, Jan Kovanda**, Petr Krejčí***

Biomechanical simulation activities are seen to undergo considerable growth in volume and scope. More complex and more complete models are now being generated. Biomechanical simulations are considered and extended well into the fields of transport vehicle occupant safety, biomedicine and virtual surgery, ergonomics and into the fields of leisure and sports article manufacture.

For an impact application like a car to pedestrian impact, correct modeling of a knee joint is important for description of the global response and dynamics after the impact. It is also useful for description of possible injuries. Based on the large research of available sources done in [3] in order to create an adequate knee joint, a simple articulated rigid body knee model is introduced. The model is based on the nonlinear joint accommodating flexion-extension and lateral rotation and translation. Joint characteristics are based on public experimental data. Dynamical validation of the new model is provided. The model is implemented into existing human articulated rigid body model ROBBY2 [2] and the frontal impact of a van versus a pedestrian is simulated including comparison to experiment.

The pre-crash activity of the human body is also essential from the point of influencing the global body motion. Hence, the influence of active muscles on the impact kinematics is investigated and comparison to passive model is provided.

Key words : *human model, articulated, knee, pre-crash activity, impact, validation*

1. Introduction

Traffic accidents seems to be a serious problem. In case of a pedestrian versus vehicle impact, the pedestrian generally faces the consequences. This paper brings an application examples of simulation of the impact situation which is the frontal impact of a van versus a pedestrian. The situation is simulated in the PAM environment where the human body model ROBBY2 is implemented. Obatined results are compared to experiment.

The ROBBY2 articulated rigid body based model [5] is being developed and used for industrial applications since 1997. The used ROBBY2 model also contains the majority of skeletal muscle represented by bar elements. Muscle properties are defined according to the Hill-type muscle model [14]. The muscles can be used in such mode that they respect only passive properties or in the active mode. In the passive mode only the passive forces rising in cosequence of muscle elongations contribute to the total muscle forces. Then we talk about the passive ROBBY2. In the active ROBBY2 the active muscle forces are taken into account. Implemented muscles are activated to simulate a living man. Activations are determined using the muscular balance solver which is a module of the PAM environment.

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2. Knee model

The knee joint is the largest human joint [7]. When describing the knee, four bones and their articulations should be discussed: the femur, the tibia, the fibula, and the patella. The contact areas of these bones are covered by cartilages. Ligaments of the joint are the patellar ligament, tibial and fibular collateral ligament, anterior and posterior cruciate ligament, transverse ligament and coronary ligaments. The knee joint menisci have important functions as stabilization of the knee joint and load bearing. They are frequently injured. Using simple rigid body access, the response of all above-mentioned parts should be taken efficiently into account on the one hand and on the other hand, the model must not be complicated much to save the computational cost. The knee model implementation is based on the large literature review [3].

2.1. Implementation

The simple description can be provided using a general joint element in the PAM environment [14]. The general joint element has 6 degrees of freedom that can be switched on or off. All tissues acting in the knee enable flexion/extension, lateral bending and lateral shearing. Such data has been found on internet based on literature review [11]. The data are based on experiments realized at the University of Virginia and they provides lateral characteristics of the knee joint including rotation and displacement scaled to the 50 % human body, see Figure 1 and Figure 2. The dotted line shows the original stiffness of the knee. It corresponds to the fixed lateral bending and fixed lateral shearing by applying high stiffness slope. There is poor rotation in transversal plane and poor translation in medial plane for both the original and the improved model.

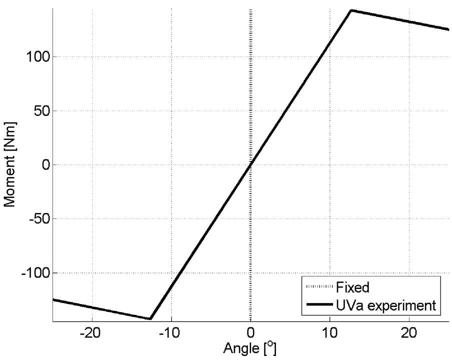

Fig.1: Knee joint characteristic curves [11] – lateral bending

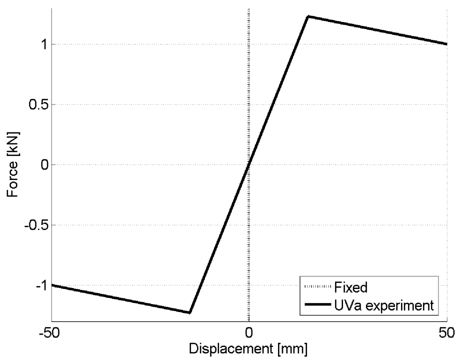

Fig.2: Knee joint characteristic curves [11] – lateral shearing

2.2. Validation

After the implementation of the new improved knee model, the ROBBY2 model [5], had to be validated to show the proper performance. The validation is based on [10] including validation for lateral bending and lateral shearing. The major problem has been due to a styrodur material that has been used as an impact face to avoid local injuries during the experiment by [10]. The simple styrodur material model to cover the impact surface has been developed [3] using normalized mechanical data [2].

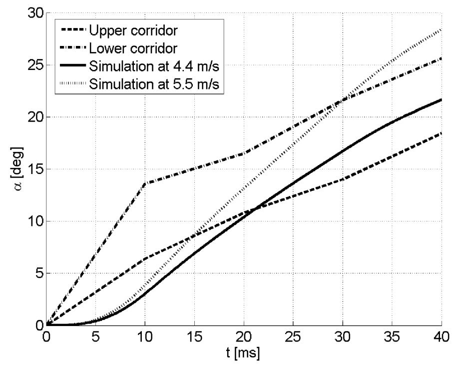

Fig.3: Bending tests – lateral bending

2.2.1. Lateral bending

Bending impacts based on [10] are performed by loading the leg just above the ankle joint. The impactor has been equipped with a foam-padded face of 50 mm thick styrodur of size 150 mm×50 mm. The distance between the knee joint line and the impactor axis on the one hand and between the knee joint line and the lower plate on the other hand have been recorded before the test.

The impact tests are performed at two impact velocities, namely 4.4 m/s and 5.5 m/s. The experimental results are scaled to the 50 % human subject. The lateral bending of the knee as well as the impactor force and the lower knee reaction force are computed and compared to the experimental data [10].

Knee flexion angle measured is similar for both velocities, see Figure 3. Impactor forces are shown in Figure 4 and Figure 5 and lower reaction that is the force between the thigh and the lower support forces are shown in Figures 6 and Figure 7. Chosen kinematic states during the impact are shown in Figure 8, Figure 9 and Figure 10.

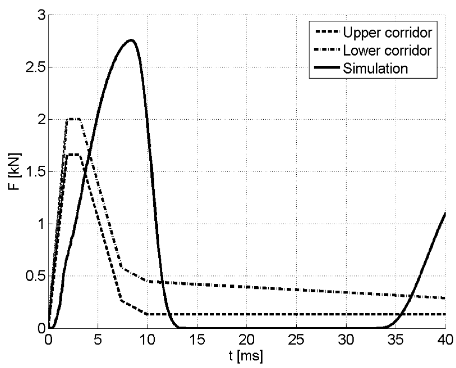

Fig.4: Bending tests at 4.4 m/s – impactor force

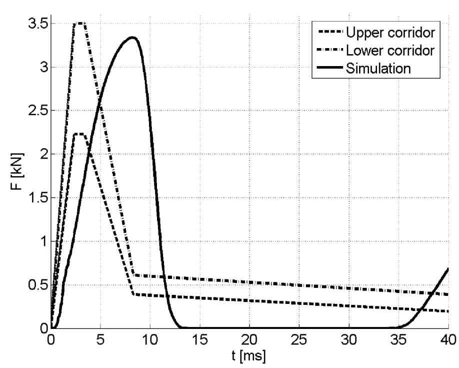

Fig.5: Bending tests at 5.5 m/s – impactor force

The knee angle shows a good trend during the simulation, however, the impactor force is higher for 4.4 m/s. The peak impactor force for 5.5 m/s is in good correlation, however it is moved to the right comparing to the experimental data. Lower reaction forces have a good trend concerning shape. For 4.4 m/s, the force is going over the experimental corridors. The

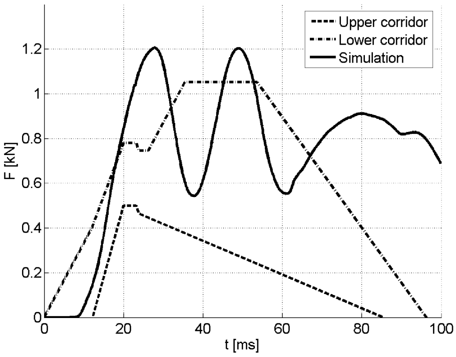

Fig.6: Bending tests at 4.4 m/s – lower reaction force

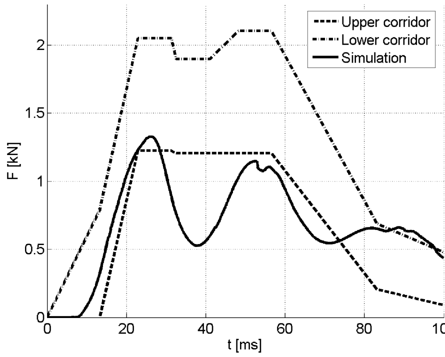

Fig.7: Bending tests at 5.5 m/s – lower reaction force

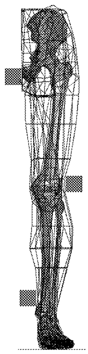

Fig.8: Bending tests – initial position

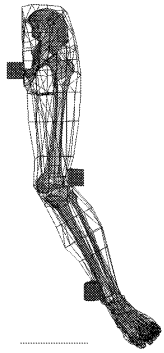

Fig.9: Bending tests at 4.4 m/s – position at 50 ms

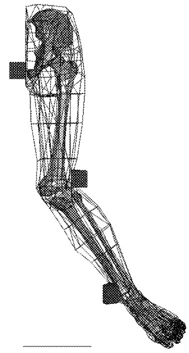

Fig.10: Bending tests at 5.5 m/s – position at 50 ms

experimental corridors are the upper and lower limits derived from the set of experimental results. On the other hand, for the 5.5 m/s, the force is under the corridors. The problem is probably caused especially by the simple material of styrodur and of course, by the simple model based on rigid bodies. The rigid bodies probably clear the situation of the area under the impactor force since there is no energy absorption of the soft tissue. Taking all the simplifications into account, the simple rigid body based knee model is validated for the bending test.

2.2.2. Lateral shearing

Shearing tests have been performed based on [10] by loading the leg with two impact plates fixed on the impactor, one loading the leg at the proximal end of tibia and head of fibula named the upper impact face, and one loading the leg just above the ankle joint and named the lower impact face. The distance between the lower plate and the upper impact interface has been chosen to be 40 mm. To minimize contact injuries, two foam-padded interfaces have been fixed on the plates (50 mm of styrodur).

The impact tests are performed at two impact velocities, namely 4.2 m/s and 5.5 m/s. The experimental results are scaled to the 50 % human subject. The upper impact plate force, lower impact plate force and knee reaction force are computed and compared to the experimental data [10].

Figure 15 and Figure 16 show the lower impactor force whilst Figure 13 and Figure 14 show the upper impactor force. The femur reaction force is shown in Figure 11 and Figure 12. Chosen kinematic states during the impact are shown in Figure 17, Figure 18 and Figure 19.

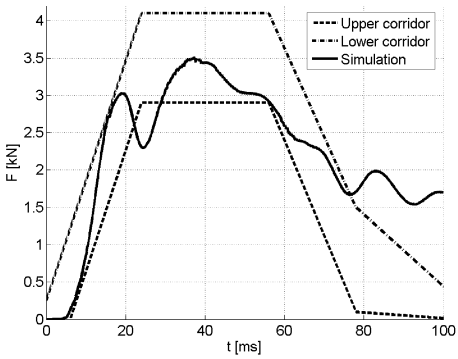

Fig.11: Shearing tests at 4.2 m/s – femur reaction force

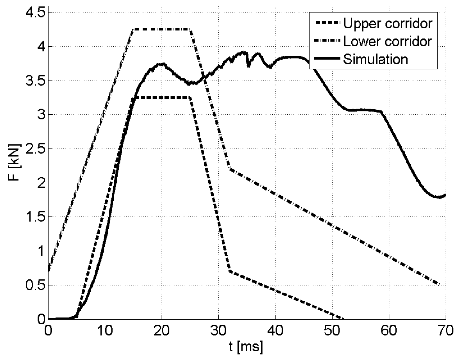

Fig.12: Shearing tests at 5.5 m/s – femur reaction force

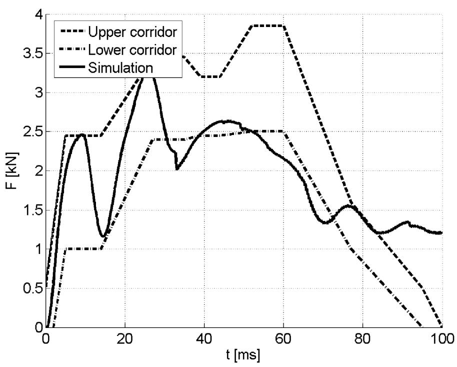

Fig.13: Shearing tests at 4.2 m/s – upper impactor force

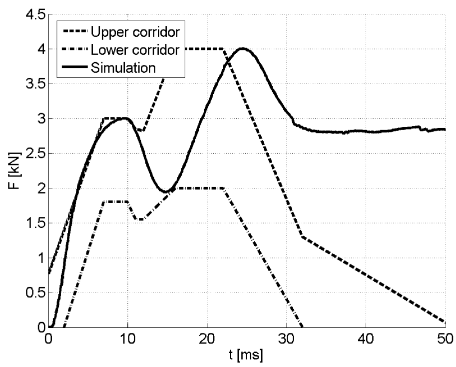

Fig.14: Shearing tests at 5.5 m/s – upper impactor force

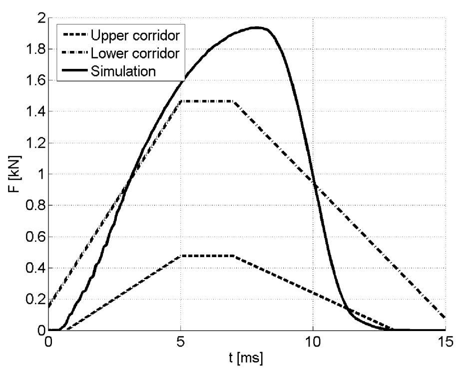

Fig.15: Shearing tests at 4.2 m/s – lower impactor force

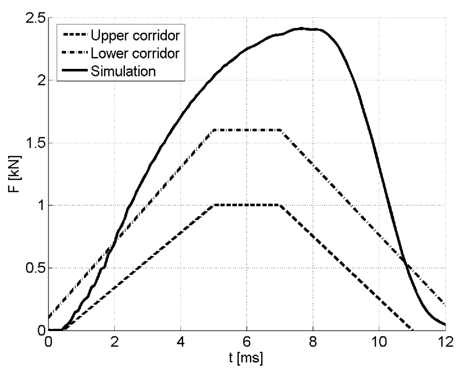

Fig.16: Shearing tests at 5.5 m/s – lower impactor force

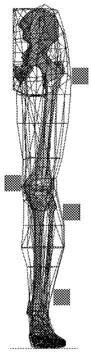

Fig.17: Shearing tests – initial position

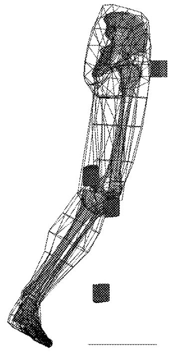

Fig.18: Shearing tests at 4.2 m/s – position at 50 ms

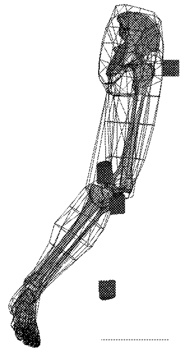

Fig.19: Shearing tests at 5.5 m/s – position at 50 ms

The lower reaction forces show very good trend concerning shape. The peak values are little bit lower. The femur reaction forces also show very good trend concerning the shape. For 4.2 m/s, the peak values are little bit lower. The upper impactor force is again over the corridor. The problem is probably caused especially by the simple material of styrodur and of course, by the simple model based on rigid bodies. The rigid bodies probably clear the situation of the area under the impactor force since there is no energy absorption of the soft tissue. Taking all the simplifications into account, the simple rigid body based knee model is validated for the shearing test.

3. Active muscle model

3.1. Material model

The muscle model in the PAM environment is defined in compliance with the Hill-type model of a skeletal muscle. The description and definitions are based on the macroscopic characteristics, which are the elongation, shortening velocity and activation degree. The Hill-type model, see Figure 20, contains the active part and the passive part. The active part represents the contractile element (CE), the passive part is the parallel visco-elastic element with the nonlinear spring (PE) and linear dashpot (DE).

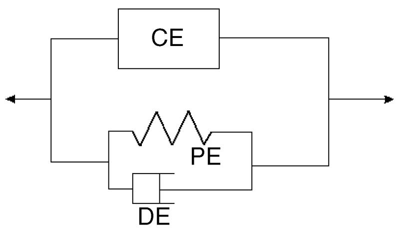

Fig.20: Hill-type muscle element model [14]

The resultant force of the muscle element is computed as the sum of forces of particular elements as

$$\mathbf{F} = \mathbf{F}_{\mathrm{CE}} + \mathbf{F}_{\mathrm{PE}} + \mathbf{F}_{\mathrm{DE}} \ . \tag{1}$$

The total muscle force-length dependecy at $v = 0$ and constant $N_{\mathrm{a}}(t)$ is displayed in Figure 21. Long muscles particularly that cross joints are modeled as a serial connection of

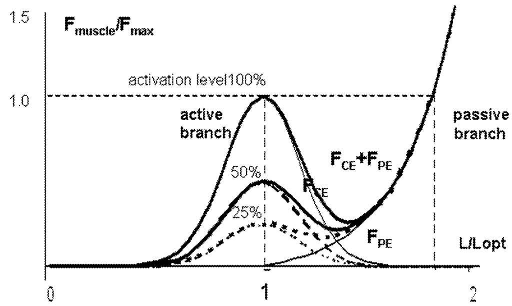

Fig.21: Dependency of the total force on length, taken from [14]

several bars. Hill-type models describe the force response of a muscle in terms of elongation of the entire anatomical muscle. Therefore the muscle forces are calculated from the sum of elongation of all elements representing a given muscle. The force produced by the contractile element is a function of its instantaneous muscle length L, its instantaneous elongation/contraction rate v, and its instantaneous value of the active muscle state $N_a(t)$ as

$$\mathbf{F}_{\mathrm{CE}}(x,v,t) = \mathbf{F}_{\max}\, N_{\mathrm{a}}(t)\, \mathbf{F}_{\mathrm{L}}(x)\, \mathbf{F}_{\mathrm{v}}(v) \ , \tag{2}$$

where $\mathbf{F}_{\mathrm{L}}(x)$ is the force-length characteristic, $\mathbf{F}_{\mathrm{v}}(v)$ is the force-velocity characteristics, $\mathbf{F}_{\max}$ corresponds to maximal isometric force and $N_{\mathrm{a}}(t)$ represents the level of muscle activation, that ranges between 0 and 1. These all characteristic functions can be user defined or the muscle behavior can be influenced by parameters which appear in relations implemented in the code. The relations can be found in [14] together with their more particular definitions.

An inactivated muscle ($N_{\mathrm{a}}(t) = 0 \Rightarrow F_{\mathrm{CE}} = 0$) has passive properties that put up a resistance during elongation. This behavior is described by passive muscle force – elongation characteristic function. For more details see [14].

3.2. Muscular balance software

ESI Group has developed the muscular balance software for determination of active muscle forces in human body models. The software is created for tasks of comfort. It is intended for solving of problems of statics. The active force of each modeled muscle is determined for each loaded static position as the set of muscle forces that will sustain the imposed position in a static equilibrium. Since the number of degrees of freedom of the skeleton is less than the number of muscle segments the overdetermined system is obtained. This system is solved by optimizing algorithm, which calculates the active muscle forces acting on the articulated skeleton by minimizing the energy under zero to full muscle activation levels and for static equilibrium. The optimization function exerts to evaluate the likely distribution of the muscle forces to be activated in agonists and synergizers in a given static position as

$$f = \left(\sum_i \gamma_i\,(\alpha_i - c) \right)^{1/2} \ . \tag{3}$$

There the sum ranges over all participating muscle segments, i, c is the activation level of voluntary muscle contraction, γ has the value 0 for the antagonists and the value 1 for the load carrying muscles that are the agonists and the synergists and finally α_i is the total activation level for the i-th muscle segment. The level of voluntary and total muscle contraction ranges from 0 % to 100 %. The function in (3) can be interpreted as the least possible overall level of muscle activation in a given position when the level of the voluntary muscle contraction is c [1]. A human subject can carry a given load in a given posture under more or less overall voluntary muscle contraction (0–100 %). The voluntary contraction can be represented as an ability of human being to willingly tense its muscles without carrying any load.

4. Pedestrian impact

The new improved knee model is implemented into the ROBBY2 model and has been applied for simulation of a pedestrian impact. This knee joint enables not only flexion-extension but also lateral bending and lateral shearing. Three different cases of the simulation are compared to the experiment. A pedestrian is represented by the passive ROBBY2, passive ROBBY2 with the improved knees and active ROBBY2 with the improved knees. The term passive model indicates deactivated muscles, only the passive muscle force may contribute to the total muscle force in consequence of muscle elongation. In the active model all implemented muscles are activated by activations computed using the muscular balance software to hold upright standing position.

It is supposed that immediately after an impact, a person losts control over its movement, even it often falls unconscious and the muscle activity decreases. The shock influence is neglected. Results are analyzed with regard to a primary impact that means the contact between the pedestrian and the vehicle. Only the primary impact is taken into account since the secondary one can also cause critical consequences from the point of view of injuries but the secondary one is rather related to road infrastructure that is not dealt with this article.

4.1. Injury criteria

Using rigid body models we are not able to directly recognize accident consequences on a human body. The measure of injury is determined by different injury criteria [12] based on body parts acceleration. Following criteria were defined for car passangers however they are used also for a pedestrian impact, see [9] and [13]. The results are computed from filtered acceleration curves, see Table 1, where the filters are chosen according to the recommendation in [14].

Criterion	Filter
HIC	CFC1000
3ms	Sae180

Tab.1: Used filters [14]

With regard to a probability of injury, head, thorax and knees are the most sensitive body parts related to pedestrian impact. The general head injury criterion (HIC) is used. It is computed from the acceleration of the head gravity center as

$$\mathrm{HIC} = \max_{0 \le t_1 \le t_2 \le T} \left(\frac{1}{t_2 - t_1} \int\limits_{t_1}^{t_2} \mathrm{acc}(t)\,\mathrm{d}t \right)^{2.5} (t_2 - t_1)\ , \tag{4}$$

where t_2 and t_1 are two arbitrary times during the acceleration pulse. The victim of an accident sustains a head injury if $\mathrm{HIC}_{36} < 1000$ according to the regulation EC102/2003, where the time window is $t_2 - t_1 \leq 36\,\mathrm{ms}$. Injuries of organs embedded in the thorax are judged according to 3 ms criterion, which is defined as the highest acceleration pulse lasting at least 3 ms. This pulse can not exceed 60 g according to the regulation EC102/2003. One possibility how to indicate the knee injury is to monitore its lateral bending angle. According to [11] it is supposed that a knee failure occurs if the lateral bending angle reaches approximately 13°.

4.2. Frontal impact

Within the scope of the FT-TA/024 project and in cooperation with Czech Technical University in Prague an experiment has been realized. A frontal impact of a pedestrian (dummy MANIKIN) versus a van (Ford Tranzit) [8]. This frontal impact has been firstly simulated in MADYMO and it has been validated with mentioned experimental data, see also [8].

The situation is simulated again in the PAM environment using the ROBBY2 model implemented in three different above-mentioned variants: the passive ROBBY2 (ROBBY2P), the passive ROBBY2 with the improoved knee joints (ROBBY2PK) and the active ROBBY2 with the improoved knee joints (ROBBY2AK). The muscle activations of the standing ROBBY2 are switched off 50 ms after impact in consequence of the supposed lost of control.

The geometry and material parameters of the van model are taken from [8]. The van model is simplified so that the vehicle body is modeled by rigid ellipsoids. The ROBBY model is placed on the rigid support frontaly to the van. The impact velocity is 28 km/h and the vehicle is breaking by the deceleration equal to 0.8 g. Real friction coefficients between the pedestrian and the ground and contact characteristics between the pedestrian and various parts of the car are used.

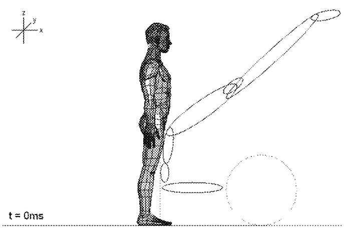

Fig.22: Frontal impact – initial position

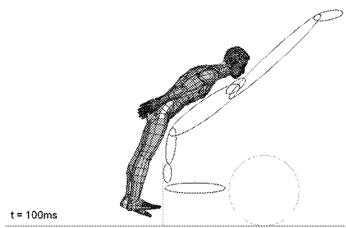

Fig.23: Frontal impact – position at 100 ms

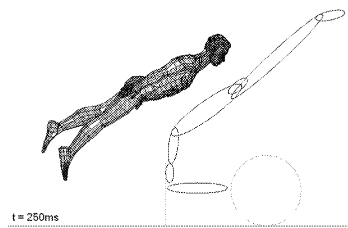

Fig.24: Frontal impact – position at 250 ms

The van firstly impacts the area of pelvis and thighs then the thorax strikes on the bonnet followed by the head. Selected simulation states are shown in Figure 22, Figure 23 and Figure 24. The head accelerations in local coordinate output compared to the experiment are shown in Figure 25, Figure 26 and Figure 27. The head acceleration magnitude compared to the experiment is shown in Figure 28. The thorax frontal acceleration compared to the experiment is shown in Figure 29. The all displayed curves are filtered by Sae60 due to a clear representation. One can see that there is no significant difference between the behavior of the original model and the model with the new improved knee since there is a small lateral motion component. However the injury criteria are influenced and the improved knee model shows better correspondance to the experimental results. As supposed, the passive ROBBY2 model is close to the MANIKIN response since MANIKIN is also passive. Only the head

Y acceleration is different from the experiment since MANIKIN contains only rotational joints hence it is used in the frontal direction. The ROBBY's head Y acceleration is almost zero since the motion (at the beginning at least) is planar. The active muscles slightly influence the model kinematics whilst the injury criteria are influenced significantly. Even the muscles are activated only in legs, the stiffening of the lower part influences the stiffening of the whole body including the head. The computed and measured HIC and 3ms criteria are compared in Table 2. The HIC criterion is computed from the head magnitude acceleration. The 3ms criterion should be also computed from thorax magnitude acceleration but it is computed from the acceleration of thorax in the frontal direction because of no experimental data of the thorax magnitude acceleration. However, the frontal acceleration is considerably higher than the other two components that is proved by simulation, so the results compared to injury limits should not be affected much. The pre-crash muscle acitivity shows to be an important issue within the human body modelling since it stiffers the body just before the impact and hence it influences the global kinematics and dynamics of the motion.

	Injury limit	Experiment	ROBBY2P	ROBBY2PK	ROBBY2AK
HIC [-]	1000	7467	6317	6868	8485
3ms [g]	60	37.2	27.0	37.4	54.8

Tab.2: Injury criteria for the frontal impact

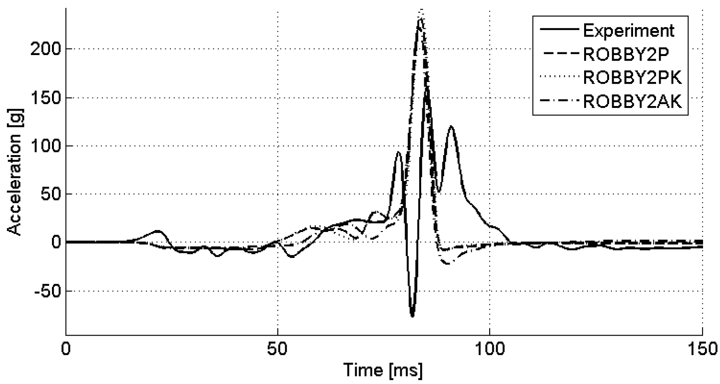

Fig.25: Head acceleration in X local coordinate output

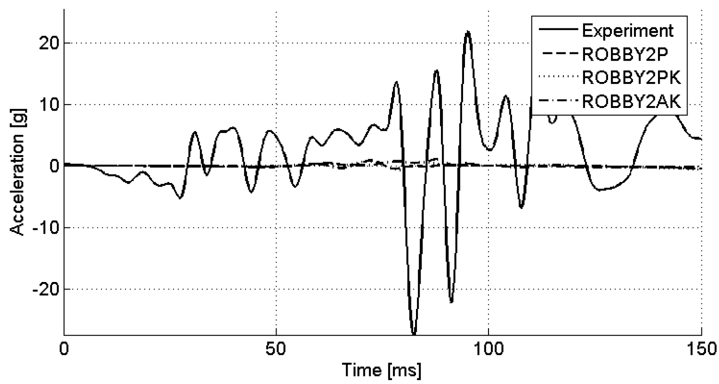

Fig.26: Head accelerationin Y local coordinate output

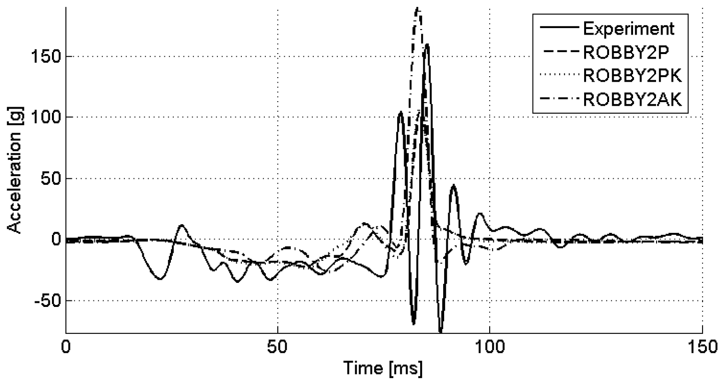

Fig.27: Head acceleration in Z local coordinate output

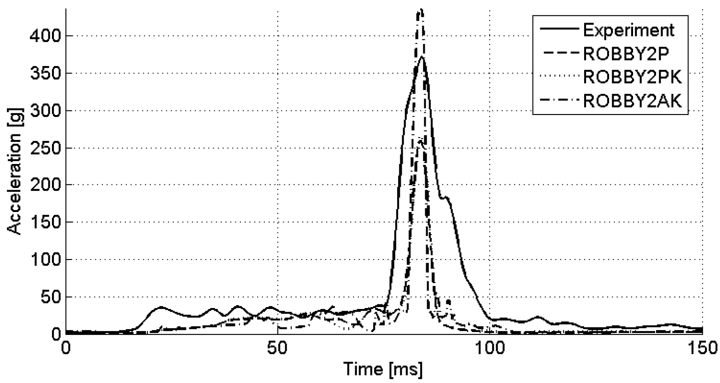

Fig.28: Head acceleration magnitude

	Left knee [deg]	Right knee [deg]
Injury limit	12.70	12.70
ROBBY2PK	7.72	7.98
ROBBY2AK	6.05	4.62

Tab.3: Maximum lateral bending angles of knees

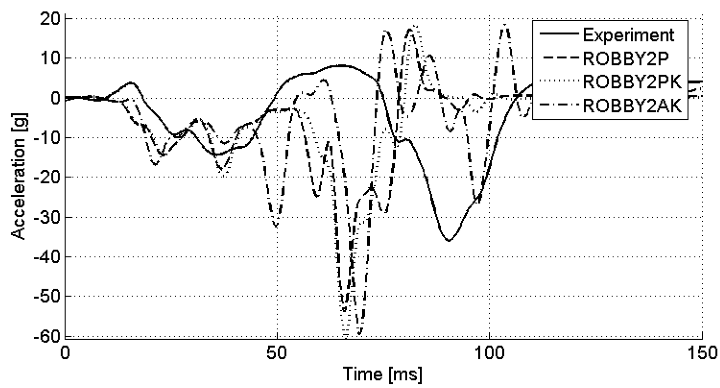

Fig.29: Thorax acceleration in X local coordinate output

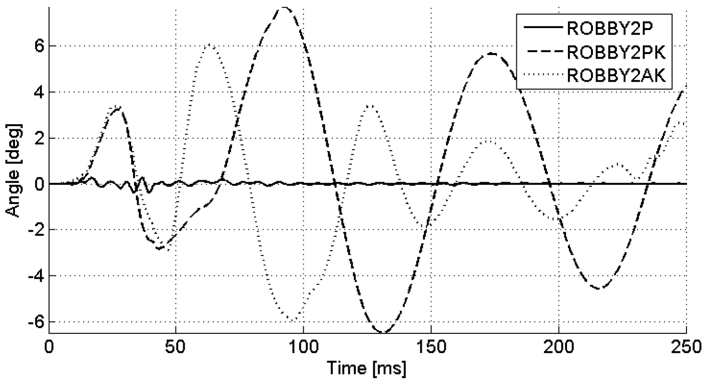

Fig.30: Left knee bending

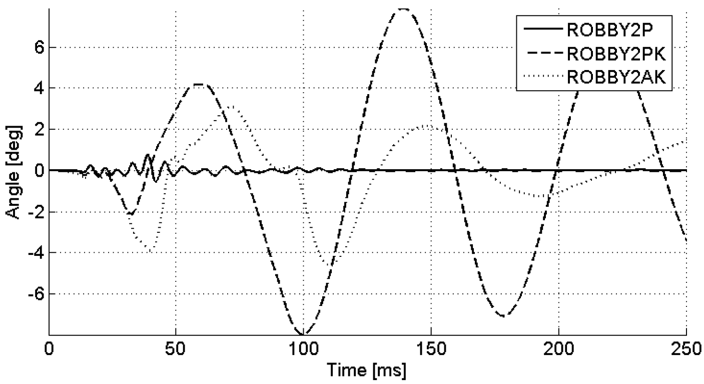

Fig.31: Right knee bending

5. Conclusion

The paper deals with the development of a simple rigid body based knee model. Based on literature review, the original rotational joint that enabled only flexion-extension has been supplied by a general three degrees of freedom joint enabling further lateral bending and lateral shearing. The response data are based on experimental data done at the University of Virginia. The new knee model was validated based on experimental data found in literature for lateral bending and lateral shearing dynamical tests. The model shows good correlation to the experimental data taking all simplifications into account.

Further the paper deals with the modeling of frontal pedestrian impacts. It is shown that the correct knee model shows different bending that influence not only the kinematics but also the dynamical characteristics. The pre-crash activity simulated by active muscles to hold the standing position before the impact is introduced and strong influence to the evaluation of injuries is proved.

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