Neck modelling for rear-end impact simulations - a comparison between a multi body system (MBS) and a finite element (FE) model

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Introduction

Soft tissue neck injuries, classified as minor (AIS1) that often do not exhibit a morphological manifestation, are on increase and of continued concern in traffic safety. In order to elucidate the kinematics as well as being able to evaluate the risk of injury, the use of mathematical modelling is well established. In the fields of impact biomechanics, mainly two different approaches of mathematical simulation techniques are used: multi body systems (MBS) and the finite element (FE) method.

The aim of this study was to subject two different neck models, a MBS and a FE model, to a range of crash pulses representing low speed rear-end impacts, all leading to the same Δv (4 m/s). The kinematic responses were compared and the ability of the models to calculate different neck injury criteria was assessed. Furthermore, the influence of the simulation method used on the results of the injury criteria was demonstrated.

Materials and Methods

METHODOLOGY -Two mathematical neck models, a MBS and a FE model, were validated against volunteer data. After that a wide range of acceleration pulses representing rear-end collisions were applied on the first thoracic vertebra (T1). For each pulse the kinematic response of the models was calculated and the neck injury criteria chosen for comparison were determined.

MULTI BODY SYSTEM -The Multi Body System neck model was developed with the software program MADYMO 2D [1], with motion restricted to the sagittal plane. The neck model consisted of seven cervical segments and one thoracic segment, modelled by means of bodies connected to each other by revolute joints (Figure 1). The total stiffness of the neck was achieved by complementing the individual joint stiffness with two string muscle substitutes: one on the posterior side and one on the anterior side of the neck. The muscle substitutes were modelled using flexible tension elements connected to elements containing both damping and elastic load characteristics [2].

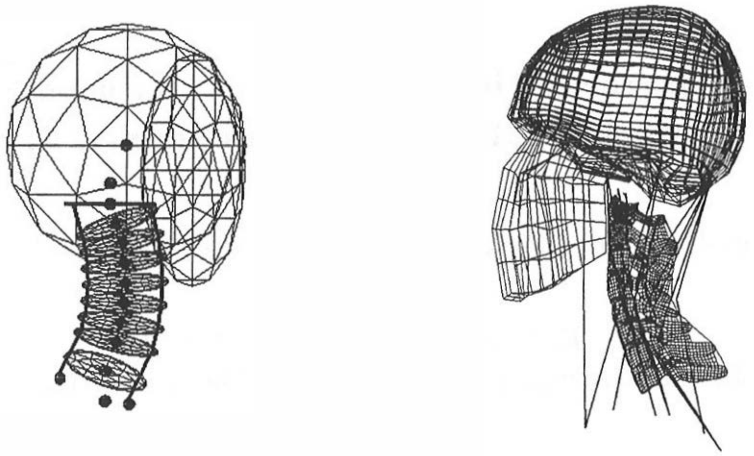

Figure 1: 2D MBS model (left) and 3D FE model (right).

FINITE ELEMENT MODEL -The FE model, a three-dimensional model of the cervical spine and the first thoracic vertebra (Figure 1), was computed with PAM-CRASH [3] software. Based on a model previously developed by Yang [4], all bony structures, the intervertebral discs, all major ligaments as well as the most relevant neck muscles were included into the model. The muscles implemented used a Hill-type formulation, thus taking into account the active and passive muscle characteristics. A reflex time of 50 ms for all muscles was used according to EMG measurement [5]. Other parameters were adopted from literature [6].

VALIDATION - Validation was performed using volunteer data obtained from the Japan Automobile Research Institute (JARI). The data was gained from the evaluation of volunteer sled test experiments using a test set-up similar to that described in [7]. The subjects were seated on a seat without head restraint; the impact speed was set to 8 km/h. The kinematics of the cervical vertebrae of the volunteers was recorded with appropriate instrumentation including cineradiography.

INPUT DATA - For the comparison of the FE and the MBS model, five different horizontal acceleration pulses were applied in the same way to the first thoracic vertebra. The shape of the crash pulses (Figure 2) was chosen from a theoretical point of view. The peak acceleration varied between 60 to 80 m/s^2 and all crash pulses corresponded to a Δv of 4 m/s. These pulses represents a level of impact severity where soft tissue neck injuries are more likely to be sustained than at the level at which the volunteer tests used here for validation have been performed.

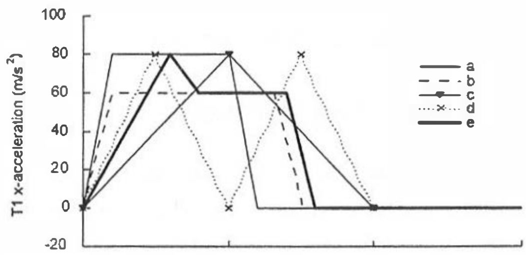

Figure 2: Acceleration pulses used as input on T1.

COMPARISON CRITERIA - The response of the different neck modelling techniques was compared in different aspects: first the computational aspects in terms of needed CPU time and memory were evaluated, next the kinematic output was analysed of which here only the relative angular displacement of the vertebrae is presented. Finally, in order to assess the ability to predict the risk of injuries by the models three injury criteria were chosen to be evaluated in this study: the neck injury criterion (NIC) [8], Nij, [9] and the intervertebral neck injury criterion (IV-NIC) [10].

Results and Discussion

The validation of the MBS and the FE model agreed reasonably well with the experimental data. For the FE model the head x-acceleration was within the corridor of the volunteer data. The peak head x-acceleration of the MBS model was 10% higher than the peak value from the corridor of the 8 volunteers when applying the x-acceleration on T1. When prescribing the motion of the T1 the peak head x acceleration was within the corridor of the volunteers. Hence the models formed a suitable basis for further investigation. Comparing the computational effort necessary for the specific techniques, as expected, a much higher CPU time was needed for the FE model (factor 30 for computing the same case on an IBM RS 6000 workstation). Also the memory required for the FE model is considerably higher.

In terms of kinematic response the relative angular displacement of the MBS (Figure 3) indicated an initial flexion motion of the C2-C3 and C3-C4 segments during the first stage of impact before changing into extension as found in volunteer studies [11]. This was not derived by the FE model where extension motion from the beginning was observed for all segments. However, the maximal rotation angle of the head was determined to be similar in amplitude and timing.

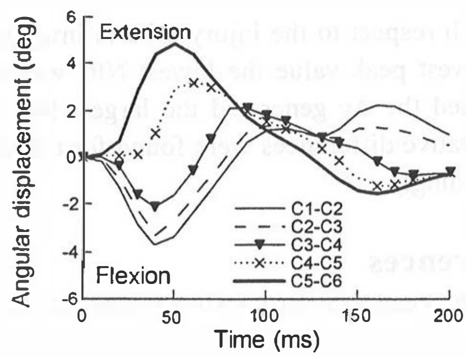

Figure 3: Angular displacement of the individual vertebrae for the Madymo neck model when crash pulse a was applied to T1.

Evaluating the injury criteria as described above, the NIC values were within a range of 9 - 16 m^2/s^2 (Table1). The values for the MBS differ slightly more than those for the FE model. This might be attributed to the underlying modelling approach, i.e. the FE model relates to a smoother response, because the coupling of the vertebrae to each other is not concentrated in one joint but linked by the intervertebral discs, ligaments, muscles and the contact surfaces of the synovial joints which depend on the chosen material and mechanical properties (e.g. damping). The pulse that obtained the lowest peak value (pulse b) did for both models produce the lowest NIC value. The pulse that most rapidly produced the Δv generated the largest NIC value. Correlation between NICmax and the maximum Δv produced before 85 ms after the crash event has for simulated sled tests been found by Eriksson and Boström [12].

Table 1: NICmax values obtained for the five different pulses applied to T1.

pulse	MBS	FE
a	14.85	15.92
b	9.10	10.53
c	10.29	12.12
d	14.39	14.61
e	11.39	13.10

Calculating the Nij criterion revealed clear differences between the two models. Both timing and the maximum value varied considerably, this is assumed to be partly caused by the way the head and neck is joined, especially in the FE model. Also it has to be noticed that this criterion is in particular very sensitive to the coincidental occurrence of the according forces and moments. Thus, slight differences in the predictions over time of forces and moments, respectively, cause significant differences for the Nij values.

Computing the IV-NIC relies solely on the rotations of the vertebrae. For crash pulses a, b, c, and e the MBS gave for all cases a value of 0.6 at approximately 50 ms in extension. The FE model values were within a range of 0.7 to 0.9 also occurring at approximately 50 ms. The results for case d were 0.5 and 0.7, respectively. However, both models predicted the maximum IV-NIC to occur in the C6-C7 segment. As to date, there is no experimental data available evaluating the IV-NIC, the application of the determined IV-NIC values can hardly be assessed in terms of injury risk.

Conclusions

Both the FE and the MBS method as represented by the models under consideration are suited for neck modelling. However, depending on the purpose of the neck model, different advantages and disadvantages have to be taken into account. Whereas the MBS model offers a good opportunity to describe the kinematic response within a short CPU time, the FE model allows a more detailed anatomical modelling and has possibilities to be combined with other fields of interest, e.g. fluid dynamics [13].

With respect to the injury criteria investigated, it can be concluded that for the pulse on T1 with the lowest peak value the lowest NIC was calculated for both models. The pulse that most rapidly produced the Δv generated the largest NIC value for both models. For the rotation based IV-NIC quantitative differences were found for the value but the maximal loaded segment was predicted correspondingly.

References

[1] TNO Road Vehicles Research Institute (1997) MADYMO User's Manual 2D. Version 5.3; Delft; The Netherlands

[2] Linder A. «A new mathematical neck model for a low-velocity rear-end impact dummy: evaluation of components influencing head kinematics»; Accident Analysis & Prevention 32; 2000; pp.261- 269

[3] ESI Paris; PAM-CRASH User's Manual Version 1998; Paris

[4] Yang K.H., Zhu F., Luan F., Zhao L. and Begeman P. »Development of a Finite Element Model of the Human Neck»; 42nd Stapp Conference; 1998; pp. 195-205

[5] Hell W., Langwieder K., Walz F, Kramer M. and Hartwig E. »Consequences for seat design due to rear end accident analysis, sled tests and possible test criteria for reducing cervical spine injuries after rear-end collision»; IRCOBI Conf.; 1999; pp. 243-259

[6] Wittek A., Kajzer J. «Modelling of muscle influence on the kinematic of the head-neck complex in impacts»; Memoirs of the School of Engineering; Nagoya University Japan; Vol. 49; No. 2;1998

[7] Ono K., Inami S., Kaneoka K, Gotou T., Kisanuki Y., Sakuma S., Miki K. »Relationship between localized spine deformation and cervical vertebral motions for low speed rear impacts using human volunteers»; IRCOBI Conf. 1999; pp. 149-164.

[8] Boström O., Svensson M., Aldman B., Hansson H., Håland Y., Lövsund P., Seeman T., Suneson A., Säljö A. and Örtengren T. »A new neck injury criterion candidate based on injury findings in the cervical spinal ganglia after experimental neck extension trauma» IRCOBI Conf.; 1996, pp. 123- 136

[9] Kleinberger M., Sun E., Eppinger R. Kuppa S., Saul R. «Development of improved injury criteria for the assessment of advance automotive restraint systems» Report of the NHTSA, September 1998.

[10] Panjabi M., Wang J., Delson N. «Neck injury criterion based on intervertebral motions and its evaluation using an instrumented neck dummy» IRCOBI Conf.; 1999; pp. 179-190

[11] Kaneoka K., Ono K., Inami S., Yokoi N., Haysshi K. "Motion analysis of cervical vertebrae during simulated whiplash loading" Proc. of the Whiplash-Associated Disorders Would Congress, Vancouver, Canada; 1999; pp. 151-160

[12] Eriksson L,; Boström O. «Assessing the influence of crash pulse, seat force characteristics, and head restraint position on NICmax in rear-end crashes using a mathematical BioRID dummy» IRCOBI Conf.; 1999; pp 213-223

[13] Schmitt K.-U., Muser M., Walz F. and Niederer P. «Biomechanics of cervical spine injuries under low speed impact conditions taking into account fluid-structure interaction» Medical & Biological Engineering & Computing; Vol.37; Suppl.2; 1999; pp.274-27