A GLOBAL HEAD NECK TORSO MODEL FOR WHIPLASH INJURY CRITERIA INVESTIGATION

Nicolas Bourdet
IMFS, University of Strasbourg
France
Rémy Willinger
IMFS, University of Strasbourg
France
Paper Number 09-0412

ABSTRACT

The development of new protective systems must be performed on tools reliable and representative of alive human. In an earlier study, a simplified but realistic modeling of the head-neck-torso system under moderate rear impact was performed. This model of minimum complexity (MC-HNT model) but able to reproduce the 5 first experimental vibration modes was validated in the frequency domain in terms of natural frequencies and damping as well as mode shapes. The human model was then coupled to a car seat-head rest complex on Madymo Code in order to give real body behaviors and accurate T1 accelerations. The hypothesis of linear behavior was used for the torso being subjected to small deformations. The present study shows in detail the methodology carried out for real-world rear impact accident reonstruction in order to establish more accurate neck injury criteria as well as associated tolerance limits. In order to proceed to that, 87 accident cases were simulated using our MC-HNT human body model coupled to 3 Toyota seats under Madymo code. Several injury criteria, such as Neck Fx, Neck Fz, T1 acceleration, NIC_{max}, N_{km} and NDC, were calculated in order to correlate the risk of AIS1 neck injury using MC-HNT. A similar work has then been done with the BioRID II model. Then a comparison between the predictive risk curves obtained by analyzing the MC-HNT model and the BioRID II model has been performed. This comparison was expressed in terms of Nagelkerke *R-square* values obtained with these analyses. It appears that the MC-HNT model gives a higher correlation than the BioRID II one for all parameter, and that the lower neck axial force is shown as the best candidate to correlate with the neck injury.

INTRODUCTION

Despite advances in safety devices, neck injuries in traffic accidents, especially non-severe rear impact accidents, are still a serious and costly social problem. The high cost of whiplash injury has been extensively documented in several countries [1,2]. In order to decrease the incidence of whiplash injuries, development of safety measures requires reliability and fidelity of human body surrogates. Most injury prevention strategies are based on impact analysis using anthropomorphic crash test dummies or mathematical models. Improvement of injury prevention techniques needs agreement between both experimental and computational models on the one hand and experimental *in vivo* human body mechanical responses on the other. Unfortunately the spine is one of the most complex structures in the human skeletal system and its behavior during impact is still poorly understood.

Today no less than three crash test dummies are used in experimental rear impact analysis: The Hybrid III dummy, developed by Foster et al [3], the BioRID II reported by Davidsson [4] and the RID dummy proposed by Cappon et al [5]. Several validation studies on neck responses have been carried out on these dummies against volunteers and post mortem subjects [4,5,6,7,8,9]. They demonstrated several limitations of this human body surrogate under low speed rear impact in terms of biofidelity. It is unclear if this lack of biofidelity is due to the torso behavior or the neck characteristics or a combination of both.

Modeling of the human trunk began in the middle of the last century and existing models can be divided into two categories i.e. continuous models [10] and lumped parameters models [11]. However, most of these models do not have a realistic behavior compared to the human body. On the one hand, models are often too detailed and involve a high number of parameters that are not easily identified with existing experimental data. On the other hand, they represent only one particular dynamic behavior of the trunk and can therefore not be used for other applications such as the simulations of rear impacts. Finally, most of the studies concerning the torso aim at characterizing the global dynamic behavior of the trunk-head system under seat ejection for military applications. In addition, none of them has studied the kinematic behavior of the first thoracic vertebra (T1) under rear impact, an essential aspect for neck injury investigation.

In previous studies undertaken by Willinger and Bourdet [12], the experimental in vivo modal analysis of the human head-neck system has provided us with natural frequencies and deformed mode shapes of this structure. Later, based on Kitazaki [13], the previous authors developed a whole human body model including the identification on five deformed mode shapes of the spine. In 2005, Bourdet *et al* [14] studied the influence of the trunk mobility.

In the first section the general methodology is presented including the use of existing experimental modal analysis for the identification of a torso lumped parameter model and its coupling to both the head neck and the car seat for rear impact applications. In the result section, the influence of trunk mobility is analyzed through comparison between responses of a rigid versus a flexible trunk under a standard rear impact pulse. Finally a parametric study is performed in order to evaluate the effect of mechanical parameters of the seat on the human neck response.

MATERIAL AND METHOD

Real-world data

In the present study the crash pulse acceleration of 87 real-life rear-end impact from Folksam database have been reconstructed. The acceleration-time history was measured during a crash by a crash pulse recorder fixed up on three car models of the same make. The recording and the analyzing have been described by [15,16,17]. The sampling rate of the crash pulse recorder is 1000 Hz during the impact phase of the crash. The acceleration data recorded were filtered at approximately 60 Hz. The occupant injury severity was divided into three categories regarding duration of symptoms; no neck injury, initial symptoms and symptoms more than one month. Examples of symptoms are neck pain, headache, dizziness and neck stiffness. The numbers of victims are presented in Table 2 for the various injury categories, car model and occupant location.

Table 1. Gender and average age for occupants with various injury categories.

	Average age	Gender (%) Male	Female
No neck injury	46	52	48
Initial symptoms	44	33	67
Symptoms > 1 month	48	47	53
Total	46	47	53

The age distribution and gender for the injury categories can be seen in Table 1. It was a similar proportion of males and females for occupants with symptoms more than one month and for all occupants. Also average age was similar for those groups.

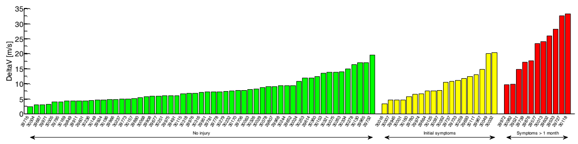


Figure 1. Representation of the Delta V of pulses versus injury severity.

Figure 1 represents the ΔV of the pulses extracted from the accident cases according to the level of injury obtained on the victim. It is interesting to observe that it appears a correlation between ΔV and injury.

Minimal complexity multi body torso model

In a previous study [12], an experimental and theoretical modal analysis of the human head-neck system in the sagittal plane have been carried out. The method has allowed to identify the mechanical properties of the head-neck system and to validate a mathematical model in the frequency domain. The extracted modal characteristics consist of a first natural frequency at 1.3 ± 0.1 Hz associated with head flexion-extension motion and a second mode at 8 ± 0.7 Hz associated with antero-posterior translation of the head, also called retraction motion, as illustrated in Figure 2.

Table 2. Number of occupants in various car models (D = Driver, FSP = Front Seat Passenger)

		Total		Car model 1		Car model 2		Car model 3	
	Total	D	FSP	D	FSP	D	FSP	D	FSP
No neck injury	77	57	20	18	4	23	11	16	5
Initial symptoms	30	19	11	2	3	13	8	4	0
Symptoms > 1 month	15	11	4	4	1	3	2	4	1
Total	122	87	35	24	8	39	21	24	6

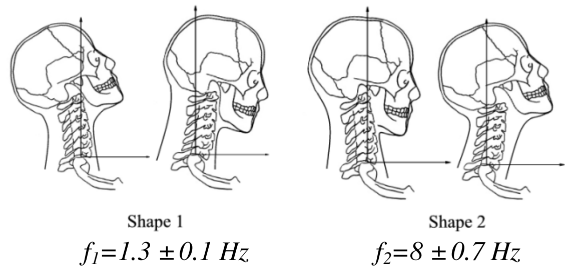


Figure 2. Representation of the two deformed mode shapes of the head neck system.

In order to address this issue, an original lumped model of the human torso was developed and coupled to a car seat-head rest complex. The hypothesis of linear behavior was used for the torso being subjected to small deformations. In a second study, the modal analysis of the human torso in a seating position conduced by Kitazaki [13] was used for both masses and mechanical properties identification [14].

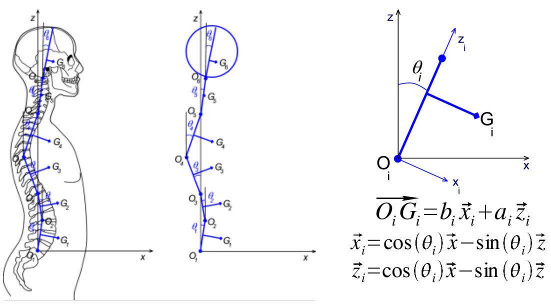
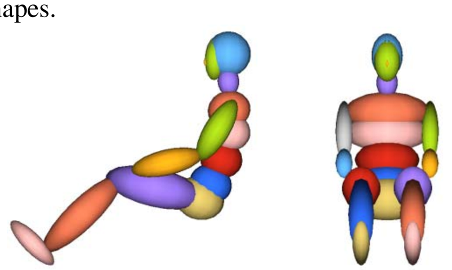


Figure 3. Representation of the lumped parameters model of the trunk, where $\theta_i=\theta_{i0}+\psi_i$ où θ_{i0} is the initial angle and ψ_i is a time dependent parameter.

In order to reproduce the four mode shapes identified experimentally the torso was divided in six segments to obtain the five degrees of freedom with the head neck system, as illustrated in Figure 3. This model of minimum complexity but able to reproduce the 5 first experimental vibration modes was validated in the frequency domain in terms of natural frequencies and damping as well as mode shapes.



(a) (b)

Figure 4. Representation of the Minimal Complexity Head Neck Torso model (MC-HNT) : (a) side view and (b) front view.

BioRID II model

To compare the MC-HNT model, the BioRID II model is used. It is the production version of a rear impact dummy developed by Chalmers University of Technology, and manufactured by Denton ATD Inc., that has been produced to meet the need for more biofidelic dummy response to rear impact events than can be obtained using a standard Hybrid III dummy. While largely based on the Hybrid III 50th percentile dummy, the BioRID II has a hinged-segment spine design, with each of the 24 vertebrae explicitly represented. Stiffening springs and dampers are fitted to model the effect of the neck muscles, and the thoracic spine and torso are more flexible than that of the Hybrid III.

Car seat models

Both models are coupled to the models of three car seats used by Kullgren *et al* [17]. In order to carry out the accident simulations, it was necessary to set the three impact configurations, i.e. to position the dummies in three seats with a torso angle of 25°. Moreover, the distance between head and headrest is defined for two configurations (50 mm and 90 mm), without modify the torso angle, in order to take account the influence of the initial seatback inclination. Figure 5 shows the setting of the dummies in various seats configurations. Moreover it shows that the initial conditions are very close between for both models.

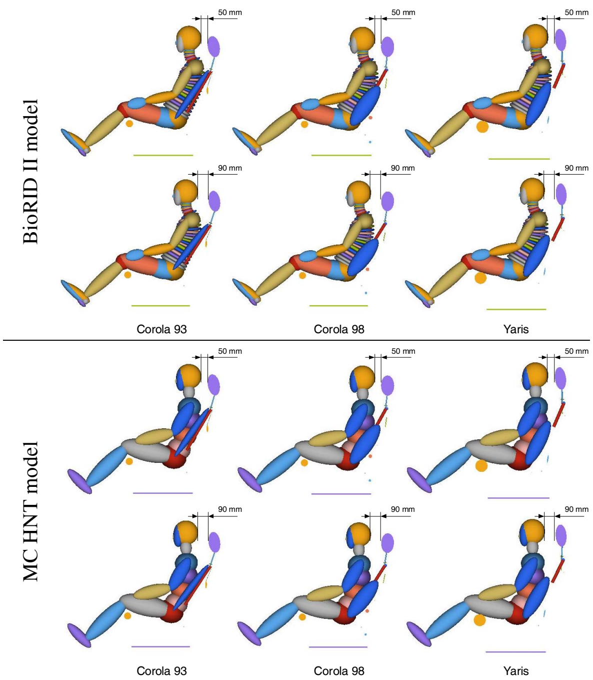


Figure 5. Position of the BioRID II and MC-HNT models in the three seats with two head-headrest distances.

87 real accidents have been reconstructed under both distance configurations and with both models. Thus, 348 simulations have been carried out. A

logistic regression has been conducted giving the Nagelkerke R^2 for the risk of initial symptom and symptoms over than one month for each model.

Accident simulations

The 87 real rear impacts accidents were simulated with both BioRID II and Minimal Complexity Head-Neck-Torso models seated on three car seats modeled in a previous study [17]. These reconstructions aim at analyzing the model behavior in order to investigate the correlation between the output parameters, as Neck Fx, Neck Fz and neck My at lower and upper neck as well as head and T1 acceleration, and three criteria as NIC_{max} [18], N_{km} [19] and NDC [20], with the injury severity.

Statistic correlation analysis

Injury correlation was evaluated by calculating the correlation coefficient, of logistic regression for each mechanical candidate parameter. The correlation coefficient R^2 proposed by Nagelkerke in 1991 [21] was used. This coefficient permits it to evaluate the quality of the regression. For that, a sample $(x_i, y_i)_{i=1,\dots,N}$ was introduced, where the x_i are the observed values of the explicative variable x and y_i are the random variable of y taking 0 for no injured and 1 for injured at case i. The logistic regression model used is a logistic function written in equation (1) which defines the probability of injury for various x.

$$p(y=1|x) = \frac{e^{\alpha+\beta x}}{1+e^{\alpha+\beta x}} \qquad (1)$$

The maximum of likelihood is calculated to identify the α and β constants. The likelihood is defined as equation (2).

$$L(\alpha,\beta) = \prod_{i=1}^{n} p(x_i)^{y_i} \left(1-p(x_i)\right)^{1-y_i} \qquad (2)$$

$$\ell(\alpha,\beta) = -\ln\left(L(\alpha,\beta)\right) \qquad (3)$$

The maximum of likelihood criterion defines the α and β values that give a maximum likelihood. For this purpose, it must be considered the opposite of its logarithm: the *log-likelihood* function defined in equation (3). This function is minimized by using the Newton Raphson's algorithm.

The correlation coefficient is well established in classical regression analysis [22]. It is defined as the proportion of variance explained by the regression model. It is used as a measure of success of predicting the dependent variable from the independent variable. In order to generalized the concept to models without easily residual variance and where the maximum of likelihood is its criterion of fit, in 1989, Cox and Snell [23] proposed a R^2 as defined in equation (4). This correlation coefficient was modified by Nagelkerke in 1991 [21], in order to "normalize" the result, expressed in equation (5), where N_k is the population number of responses in category k, i.e. 0 or 1.

$$R_{RC}^2 = 1 - \left(\frac{L(\alpha_0,0)}{L(\alpha,\beta)}\right)^{2/N} \qquad (4)$$

$$R_N^2 = \frac{R_{RC}^2}{1-L(\alpha_0,0)^{2/N}} \qquad (5)$$

$$\text{With } \alpha_0 = \ln\left(\frac{N_1}{N_0}\right) \qquad (6)$$

For each parameter, two logistic regressions are calculated: the risk of initial symptoms and the risk of symptoms over one month. The total number of occupants is 122. Concerning the risk of initial symptoms, the occupants with initial symptoms and the occupants with symptoms over one month were merged. Thus the logistic regression was calculated for 77 no-injured cases and 45 injured cases. In the same way, for the risk of symptoms over one month, the occupants with initial symptoms were regrouped with the no injured occupants. Thus the logistic regression was calculated for 107 cases having at most initial symptoms and 15 cases with long-term symptoms.

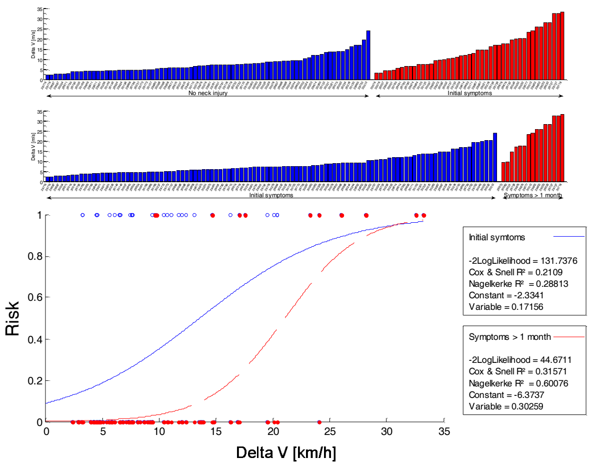


Figure 6. Histograms of Delta-V used for the two logistic regressions, and Delta-V risk curves of initial symptoms and symptoms > 1 month.

As an example, Figure 6 represents the Delta-V histogram and risk curves of initial symptoms and long term symptoms versus Delta-V for the two logistic regressions. It can be observed that Delta-V correlates well the long-term symptoms with a R^2_N of 0.6. For a 50% risk of initial symptoms, the Delta-V is of 21 km/h and for 20% risk of symptoms over one month is of 15 km/h. In 2005, Krafft *et al* [24] found 18 km/h for the same risk.

RESULTS

Comparative results on one case

An example of a comparative accident reconstruction is shown in detail in Figure 7, for a distance between head and headrest of 50 mm. It illustrates the behavior differences of the head-neck system between the BioRID II Madymo model and the MC-HNT model. Indeed, BioRID II has a light translation movement of the head followed at 150 milliseconds by an extension movement, while the MC-HNT model continues its retraction motion until 200 milliseconds.

Figure 8 and Figure 9 represent T_1 linear accelerations according to X and Z axis. In spite of prevalent oscillations on MC-HNT model, the behavior is coarsely identical. The behavior difference is especially illustrated in Figure 10 which represents the rotation of the first thoracic vertebrae (T1). Indeed, while the BioRID II upper thorax rotates forward (positive rotation), the MC-HNT model's ones undergoes an extension. Figure 11 shows relative rotation between the head and the neck for both models. It can be observed that BioRID II does not present any retraction until 120 milliseconds (positive relative rotation), while the MC-HNT model presents this movement clearly at approximately 110 milliseconds.

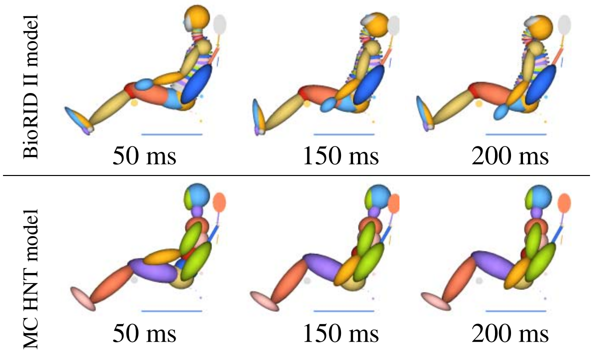


Figure 7. Pictures extracted from a simulation of case for the BioRID II and MC-HNT models.

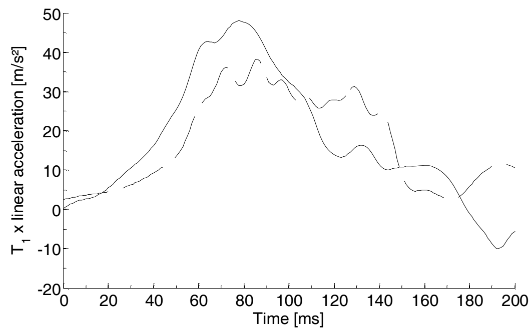


Figure 8. Superimposition of T_1 x linear acceleration for both models (— BioRID II model, – – MC-HNT model)

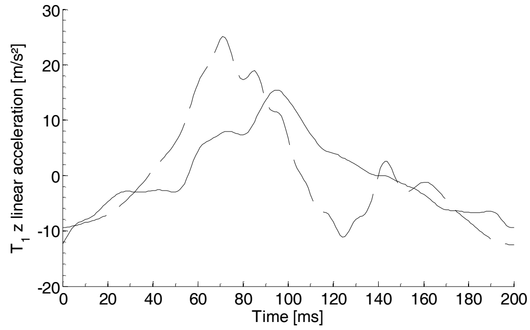


Figure 9. Superimposition of T_1 z linear acceleration for both models (— BioRID II model, – – MC-HNT model)

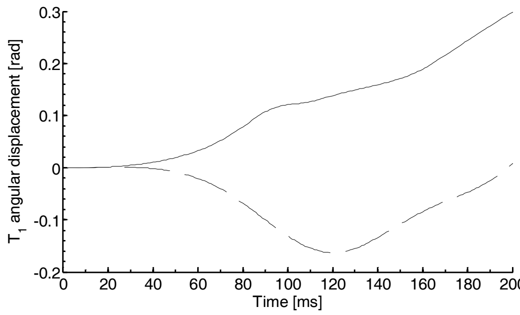


Figure 10. Superimposition of T1 angular displacement for both models (— BioRID II model, – – MC-HNT model)

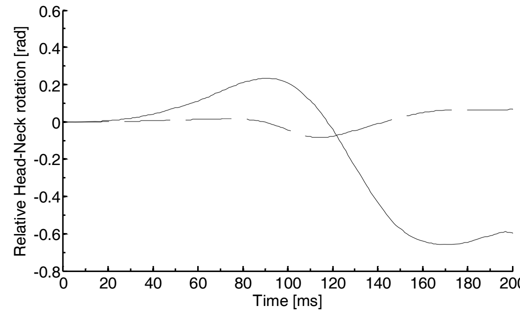


Figure 11. Superimposition of Head angular displacement both models (— BioRID II model, – – MC-HNT model)

Global statistic correlation

Figure 13 to Figure 14 represent the Nagelkerke's *R-squared* of each parameter under the two configurations for each model. It is clear that the results from the BioRID II models give different regressions according the distance between head and headrest. Indeed, concerning the risk of initial symptom, on one configuration the best candidates are T1 acceleration and the upper neck axial force as well as the lower neck shear force with R_N^2 of 0.26, R_N^2 of 0.25 and R_N^2 of 0.25 respectively. On the other configuration the best correlation is obtained with the upper neck axial force with R_N^2 of 0.10. In contrast, for the MC-HNT model, the best candidates tied are the lower neck moment, shear and axial forces as well as the upper neck axial force with a Nagelkerke *R-squared* of 0.3 under the shorter distance head-headrest configuration, and 0.32 under the higher distance configuration, adding the upper neck shear force and N_{km} criterion.

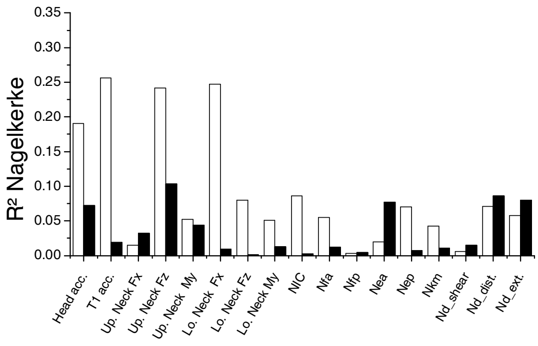


Figure 12. BioRID II model R^2_N for the risk of initial symptom (□ $d_{headrest}$=50 mm, ■ $d_{headrest}$=90 mm).

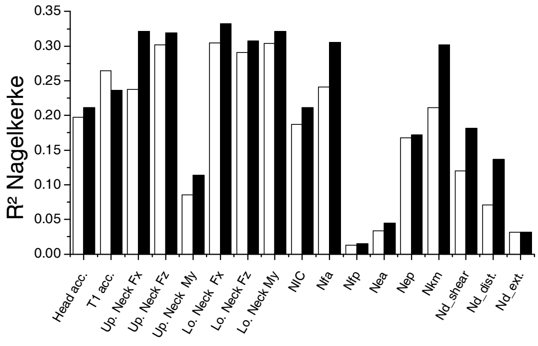


Figure 13. MC-HNT model R^2_N for the risk of initial symptom (□ $d_{headrest}$=50 mm, ■ $d_{headrest}$=90 mm).

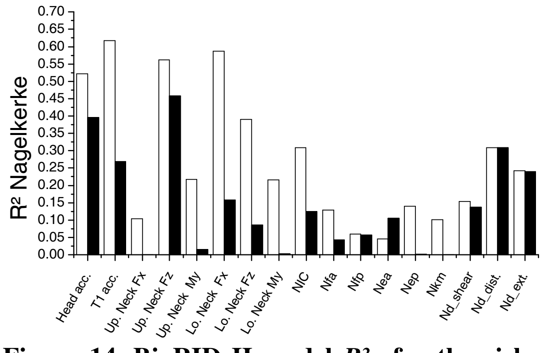


Figure 14. BioRID II model R^2_N for the risk of symptom > 1 month (□ $d_{headrest}$ = 50 mm, ■ $d_{headrest}$ = 90 mm).

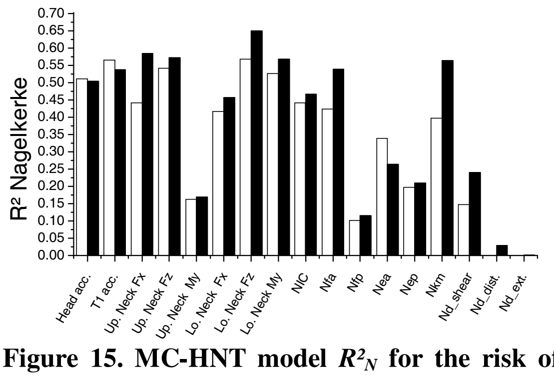


Figure 15. MC-HNT model R^2_N for the risk of symptom > 1 month (□ $d_{headrest}$ = 50 mm, ■ $d_{headrest}$ = 90 mm).

Concerning the risk of symptoms over one month, the BioRID II model shows various best candidate parameters according to the configuration. For instance, the best candidates for the first configuration are the T1 acceleration, the lower neck shear force and the upper neck axial force with $R^2_N = 0.62$ $R^2_N = 0.59$ and $R^2_N = 0.56$ respectively, and for the other configuration the best ones are the upper neck axial force and the head acceleration with $R^2_N = 0.46$ and $R^2_N = 0.40$ respectively. On the contrary, the MC-HNT model is better correlated by the lower and upper neck axial force ($R^2_N = 0.57$ and $R^2_N = 0.54$ respectively) as well as T1 acceleration with $R^2_N = 0.56$ in the first configuration case. The lower neck axial force is clearly the best candidate in case of distance head-headrest of 90 mm with $R^2_N = 0.65$.

No clear correlation between the common injury criteria and the injury outcome could be found for BioRID II model, as shown in Figure 12 and Figure 14. The best scores is given by the NIC with $R^2_N = 0.09$ concerning the risk of initial symptoms and $R^2_N = 0.31$ for the risk of symptoms over one month in case of shorter distance configuration, and $Nd_{distraction}$ in case of higher distance configuration with $R^2_N = 0.09$ and $R^2_N = 0.31$ respectively.

Candidate parameters to injury correlation

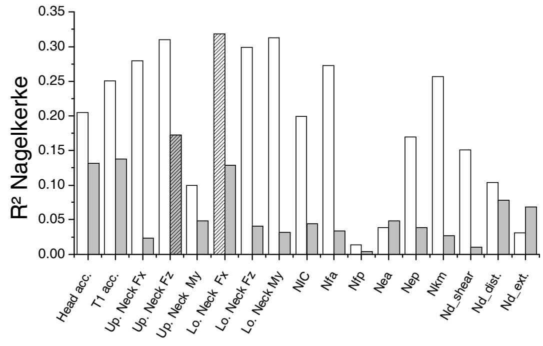


Figure 16. Representation of the mean R^2_N obtained with both model and the maximum value in black, for initial symptom (□ MC-HNT, ▒ BioRID II, ▨ Maximum value).

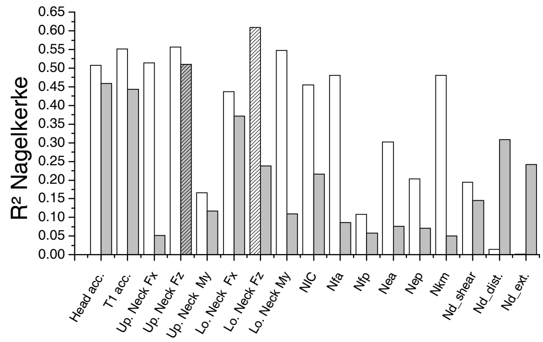


Figure 17. Representation of the mean R^2_N obtained with both and the maximum value in black model for symptom > 1 month (□ MC-HNT, ▒ BioRID II, ▨ Maximum value).

In order to extract the parameter which seems to present the correlation with neck symptoms, a mean value of R^2_N was calculated from the two configurations values. Thus, Figure 16 and Figure 17 gather the MC-HNT and the BioRID scores highlighting the best candidate represented by hatched histograms.

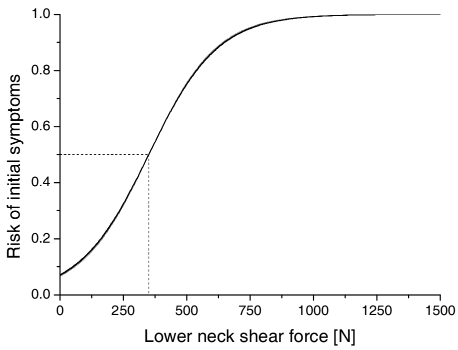


Figure 18. Lower neck shear force risk curve of initial symptoms. Grey area represents risks limited by the minimum and the maximum criterion values for both distance configuration and black line is the median (MC-HNT model).

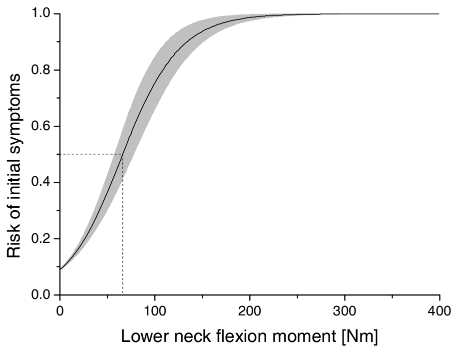


Figure 19. Lower neck moment risk curve of initial symptoms. Grey area represents risks limited by the minimum and the maximum criterion values for both distance configuration and black line is the median (MC-HNT model).

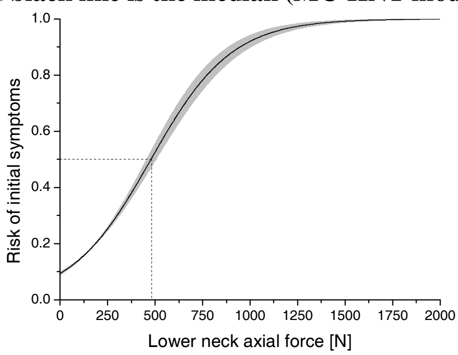


Figure 20. Lower neck axial force risk curve of initial symptoms. Grey area represents risks limited by the minimum and the maximum criterion values for both distance configuration and black line is the median (MC-HNT model).

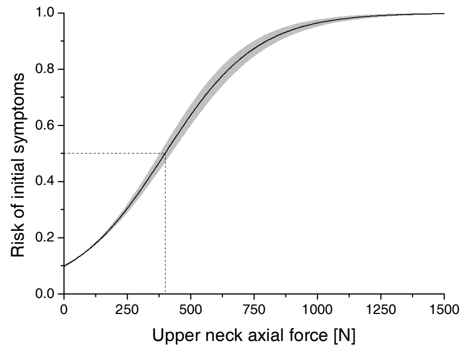


Figure 21. Upper neck axial force risk curve of initial symptoms. Grey area represents risks limited by the minimum and the maximum criterion values for both distance configuration and black line is the median (MC-HNT model).

For MC-HNT model, the lower neck shear force seems to be the parameter which give a best regression to evaluate the risk of initial symptoms with R^2_{Nmean}= 0.31. We can observe that, the lower neck moment and axial force as well as the upper axial force give R^2_{Nmean} very close to the maximum. For BioRID II model, the best parameter candidate is the upper neck axial force with $R^2_{Nmean} = 0.17$. Figure 18 to Figure 21 represent the risk curve of the four best candidate parameters. The limit at 50% risk of initial symptoms is about 68 ± 9 Nm for the lower neck flexion moment, 480 ± 24 N for the lower neck axial force, 350 ± 0.01 N for the lower neck shear force and 400 ± 20 N for the upper neck axial force. The risk curves obtained from both distance configurations are very close themselves. Except for the lower neck moment, the deviations of the limit at 50% risk don't exceed 10%.

In contrast, for the BioRID II model, the four best candidate parameters are the upper neck axial force, T1 and Head acceleration and the lower neck shear force with R^2_{Nmean}= 0.17, R^2_{Nmean}= 0.14, R^2_{Nmean} = 0.13 and R^2_{Nmean}= 0.13 respectively.

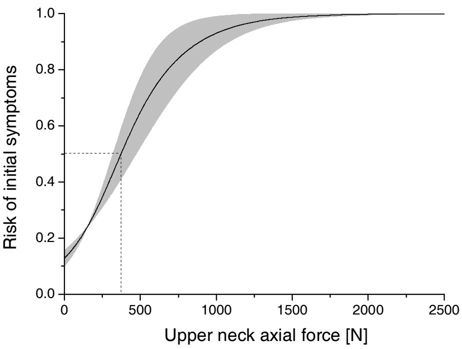


Figure 22. Upper neck axial force risk curve of initial symptoms. Grey area represents risks limited by the minimum and the maximum criterion values for both distance configuration and black line is the median (BioRID II model).

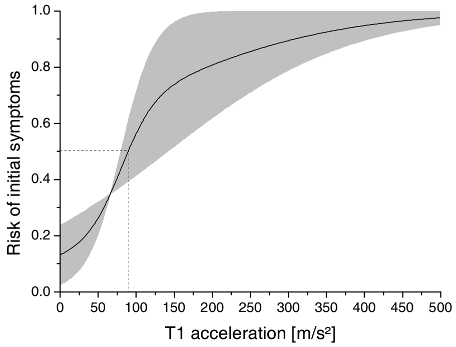


Figure 23. T1 acceleration risk curve of initial symptoms. Grey area represents risks limited by the minimum and the maximum criterion values for both distance configuration and black line is the median (BioRID II model).

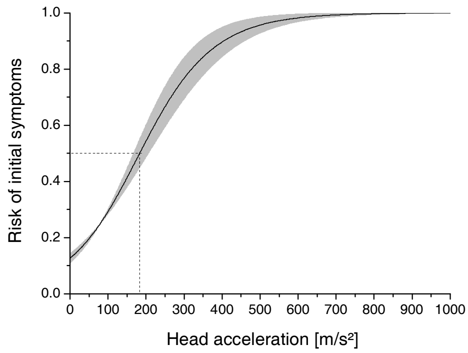


Figure 24. Head acceleration risk curve of initial symptoms. Grey area represents risks limited by the minimum and the maximum criterion values for both distance configuration and black line is the median (BioRID II model).

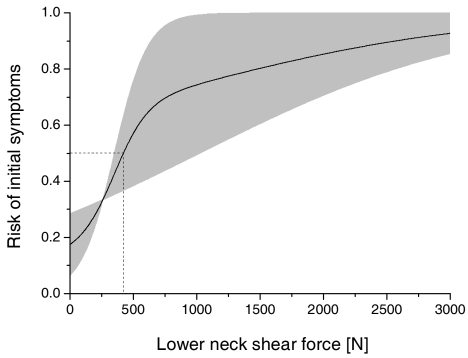


Figure 25. Lower neck shear force risk curve of initial symptoms. Grey area represents risks limited by the minimum and the maximum criterion values for both distance configuration and black line is the median (BioRID II model).

The BioRID II model risk curves of the best regressions define large corridors. The limit values at 50% risk of initial symptoms are 373 ± 75 N for the upper neck axial force, 9 ± 3 g for T1 acceleration, 18.4 ± 2 g for Head acceleration and 422 ± 336 N for the lower neck shear force. The deviations of the limits exceed 40%, as illustrated in Figure 23 to Figure 25.

Regarding the risk of symptoms over one month, Figure 26 represents the risk curves for the lower neck axial force obtained from MC-HNT model. As for the initial symptoms, the curves are very close giving a 50% risk of 703 ± 28 N.

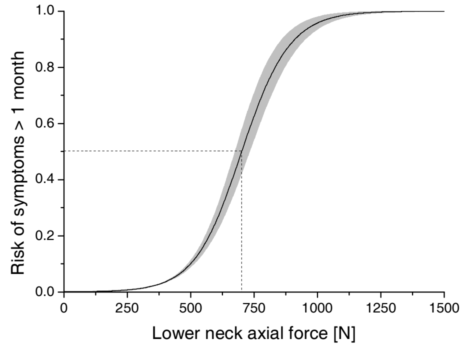


Figure 26. Lower neck axial force risk curve of initial symptoms. Grey area represents risks limited by the minimum and the maximum criterion values for both distance configuration and black line is the median (MC-HNT model).

For BioRID II, the risk at 50% of symptoms over than one month for the upper neck axial force is 516 ± 38 N, as shown in Figure 27. The deviation is smaller than for initial symptoms risk (15% against 40 % for initial symptoms risk).

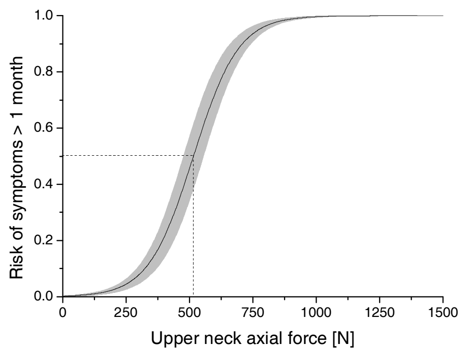


Figure 27. Upper neck axial force risk curve of initial symptoms. Grey area represents risks limited by the minimum and the maximum criterion values for both distance configuration and black line is the median (BioRID II model).

DISCUSSION

Since over ten years many investigation on new neck injury criteria for rear end impact have been carried out. Bolström *et al* [25] proposed the NIC (Neck injury Criterion) as a value to correlate the head-neck movement with the ganglia caused by transient pressures changes in spinal canal. It addresses the relative acceleration between head and torso in the head translational motion. The threshold proposed is of 15 m²/s². In 2002, Schmitt *et al* [19] proposed the N_{km} criterion based on the linear combination of shear force and bending moment at the occipital condyle.

In 2006, Eriksson and Kullgren [26] simulated 79 real accident cases from the same database used on our study under 100 posture of BioRID II and proposed a NIC risk curve for symptoms > 1 month. It estimates that a NIC of 24.5 ± 10 m²/s² corresponds to risk of 50 %, as illustrated in Figure 28. As for the NIC_{max}, they established a N_{km} risk curve for symptoms > 1 month. It estimates that a N_{km} of 0.5 ± 0.3 corresponding to risk of 50 %, as illustrated in Figure 29.

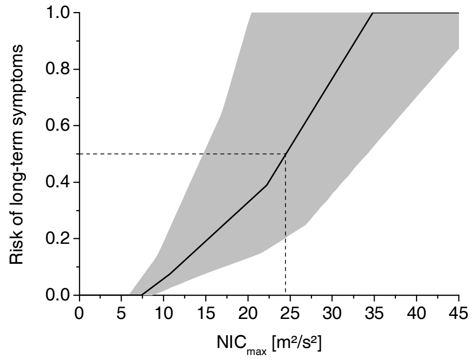


Figure 28. NIC_{max} Risk curve for symptoms > 1 month from Eriksson and Kullgren [26].

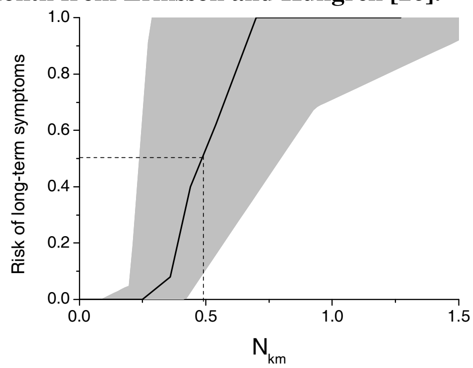


Figure 29. Risk curve of symptoms > 1 month versus lower neck axial force for both models.

Moreover, several multi body human models have been developed for rear-end impacts. A two-dimensional human model has been proposed by Jernström *et al* [27] and the computed head-torso relative angle was compared to the one recorded on a volunteer undergoing a velocity change of 8 km/h. In Jakobsson *et al* [28] it was then shown that neither the upper thoracic spine curvature of the model nor the calculated duration of the contact between the head and the headrest were in accordance with experimental data. These numerical and physical spine models are usually validated against experiments on volunteers or post-mortem human subjects (PMHS) in the time domain by superimposing model and human response parameters as a function of time. This methodology is limited as it is very difficult to characterize a multiple degrees of freedom system under impact in the time domain. The mentioned limitation illustrates the need for further torso experimental and theoretical analysis.

The present paper is to refer on in vivo human trunk characterization available in the literature using modal analysis techniques and to develop a lumped parameters model of this segment in the sagittal plane to be validated in the frequency domain. This model was seated in three car seats and 87 real accidents have been simulated. The same work has been carried out with BioRID II model in order to compare the prediction of neck damages. The simulations showed different behaviors about the head-neck-trunk system. Indeed, the MC-HNT model translation phase is longer period of time than the BioRID II model. This can be explained by the fact that the MC-HNT model presents a lower stiffness at the head-neck-trunk system. This behavior softer leads the model to be more sensitive. Indeed, during the deceleration phases, the trunk is pushed by the backseat leaving the head. The loading at the head-neck and neck-thorax junctions increases considerably if the head is pulled by the rest of the body before it contacts the headrest. This force can be projected either to axial force or to shear force according the head-thorax angle.

One of main limitations of this study is not to know the initial posture of the occupants. Indeed, the posture has a drastic influence on the behavior, leading to different logistic regressions. A first evaluation of this influence has been carried out but it should be interesting to make in deep a parameter study to extract the best candidate parameters which correlate the injury severity. Nevertheless, it highlighted the difference behavior between the two models, and the homogeneous results in terms of parameter criteria, for the MC-HNT model.

CONCLUSION

Performing 87 real accident cases from Folksam database on three seats under two configuration of distance between head and headrest using a minimum complexity model (MC-HNT model) based on the reproducibility of the 5 first experimental vibration modes of the vertebrae column revealed several parameters with higher correlation coefficient values in the logistic regression against the lesion severity. The lower neck axial force is shown as the best value of Nagelkerke R-square for both initial symptoms risk and symptoms over than one month risk.

The 87 accident cases were also simulated using BioRID II model. Then we performed a comparison between the predictive risk curves obtained by analyzing the MC-HNT model and the BioRID II model. In addition we compared the Nagelkerke R-square values obtained with these analyses. It appears that the MC-HNT model gives a better regression than the BioRID II one for all parameter.

ACKNOLEDGEMENTS

Thanks go to Anders Kullgren from Folksam Research and Karolinska Institutet of Stockholm and to Ola Boström from AUTOLIV Research.

REFERENCES

1 Szabo T.J., and Welcher J.B., Human Subject kinematics and electromyographic activity during low speed rear impacts. Proc. of the 40th Stapp Car Crash Conference, 1 Vol. SAE 962432, Albuquerque (USA), 1996, pp. 339-356.

2 Szabo T. J., Voss D. P., and Welcher J. B., Influence of seat foam and geometrical properties on BioRID P3 kinematic response to rear impacts, Proc. IRCOBI Conference, Munich (Germany), 1 Vol. pp. 87-101., 2002.

3 Foster, J.K., Kortge, J.O., and Wolanin, M.J., Hybrid III-a biomechanically-based crash test dummy, Proc. of the 21st Stapp Car Crash Conference, 1 Vol. SAE 770938, pp.975-1014, New York (USA), 1977.

4 Davidsson J., BioRID II final report, Crash Safety Division, Departement of Machine and Vehicle Design, Chalmers University of Technology, Göteborg, Sweden, 1999.

5 Cappon H., Philippens M., Ratingen v., and Wismans J. Development and evaluation of a new rear-impact crash dummy: the RID2, Proc. of the 45th Stapp Car Crash Conference, 1 Vol. SAE 2001-22-0010, pp. 225-238, San Antonio (USA), 2001.

6 Seemann M. R., Muzzy W. H., and Lustick L. S., Comparison of human and Hybrid III head and neck response, Proc. of the 30th Stapp Car Crash Conference, 1 Vol. SAE 861892, pp. 291-312, USA, 1986.

7 Prasad P., Kim A., and Weerappuli D. P. V., Biofidelity of anthropomorphic test devices for rear impact, Proc. of the 41th Stapp Car Crash Conference, 1 Vol. SAE 973342, pp. 387-415, USA, 1997.

8 Davidsson, J., Lövsund, P., Ono, K., Svensson, M., and Inami, S., A comparison between volunteer, BioRID P3 and Hybrid III performance in rear impact., Proc. IRCOBI Conference, Sitges (Spain), 1 Vol. pp. 165-190, 1999.

9 Siegmund G. P., Heinrichs B. E., Lawrence J. M., and Philippens M., Kinetic and kinematic responses of the RID2a, Hybrid III and Human Volunteers in low-speed rear-end collisions, Proc. of the 45th Stapp Car Crash Conference, 1 Vol. SAE 2001-22-0011, pp. 239-256, San Antonio (USA), 2001.

10 Hess, J.L., and Lombard, C.F., Theoretical investigations of dynamic response of man to high vertical acceleration. Aviat. Med. 29, 1958.

11 Vulcan, A.P., King, A.I., and Nakamura, G.S., Effects of bending on the vertebral column during +Gz acceleration. Aerosp. Med. 41: 294, 1970.

12 Willinger R., Bourdet N., Fischer R., Le Gall F., New method for biofidelity evaluation of dummy necks. 18th ESV Conf., 18ESV-000343, Nagoya (Japon), 2003

13 Kitazaki Satoshi, 1992, Application of experimental modal analysis to the human whole-bodyvibration. Proceedings of the United Kingdom Informal Group Meeting on Human Response to Vibration, held at The University of Southampton, Southampton, Hampshire, 28-30 September, 1992, 17-39.

14 Bourdet N., Willinger R. Human body-car seat coupling under rear impact. 19th ESV Conf., 19ESV-05-261,Washingthon (United States),2005

15 Aldman, B., Kullgren, A., Lie, A., Tingvall, C., 1991. Crash pulse recorder (CPR)-development and evaluation of a low cost device for measuring crash pulse and delta-V in real life accidents. Proc. of the 13th ESV Conf., Paris. Paper, 91-S1-W- 26, 188–192.

16 Kullgren, A., Lie, A., Tingvall, C., 1995. Crash Pulse Recorder (CPR)-validation in full scale crash tests. Accident Analysis Prevention 27, 717–727.

17 Kullgren A, Eriksson L, Boström O, Krafft M (2003) Validation of neck injury criteria usingreconstructed real-life rear-end crashes with recorded crash pulses. Proc. of the 18thTechn. Conf. on ESV, Paper No. 344, Tokyo.

18 Boström O.; Fredriksson R.; Håland Y.; Jakobsson L.; Krafft M.; Lövsund P.; Muser M.H.; Svensson M.Y.(2000) Comparison of Car Seats in low Speed Rear-End Impacts Using the BioRID Dummy and the new Neck Injury Criterion (NIC). Accident Analysis & Prevention, Volume 32, No. 2

19 Schmitt K-U, Muser M H, Walz F H, and Niederer P, Nkm— a proposal for a neck protection criterion for low speed rear-end impacts 2002, Traffic Injury Prevention, Vol. 3(2), pp. 117-126

20 Viano D, Davidsson J (2002) Neck displacements of volunteers, BioRID P3 and Hybrid III in rear impacts: implications to whiplash assessment by a neck displacement criterion (NDC), Traffic injury prevention, Vol. 3, No. 2

21 Nagelkerke N. J. D., A note on a general definition of the coefficient of determination, Biometrika, 78 (3), p. 691-692, 1991

22 Rao C.R., Linear statistical inference and its application, 2nd ed. New York, Wiley, 1973

23 Cox D.R. and Snell E.J., The analysis of binary data, 2nd ed. London, Chapman and Hall, 1989.

24 Maria Krafft, Anders Kullgren, Sigrun Malm, Anders Ydenius, Influence of crash severity on various whiplash injury symptoms: a study based on real-life rear-end crashes with recorded crash pulses, 19th ESV Conf., 19ESV-05-363,Washingthon (United States),2005

25 Boström O., Svensson M. Y., Aldman B., Hansson H., Håland Y., Lövsund P., Seeman T., Suneson A., Säljö A. and Örtengren T, (1996) "A new neck injury criterion candidate based on injury findings in the cervical spine ganglia after experimental neck extension trauma". International IRCOBI conference on the Biomechanics of Impacts, Dublin, Ireland.

26 Eriksson Linda; Kullgren Anders, Influence of Seat Geometry and Seating Posture on NIC_{max} Long-Term AIS 1 Neck Injury Predictability, Traffic Injury Prevention, Volume 7, Issue 1 March 2006 , pages 61-69.

27 Jernström C., Nilsson G. And Svensson M. Y., A first approach to an implementation in MADYMO of a human body model for rear impact modeling. Proc of the 4th International Madymo User's Meeting, september 6-7, Eindhoven (The Netherlands), 1993.

28 Jakobsson, L., Norin, H., Jernström, C., Svensson, S.-E., Johnsén, P., Isaksson-Hellman, I., and Svensson, M.Y., Analysis of Different Head and Neck Responses in Rear-End Car Collisions Using a New Humanlike Mathematical Model. Proc. of the IRCOBI Conference, Lyon (France), pp. 109-126, 1994.