Sports Engineering
https://doi.org/10.1007/s12283-018-0268-z

ORIGINAL ARTICLE

Protection from motorcycle neck–braces using FE modelling

Frank Meyer[1] · Caroline Deck[1] · Rémy Willinger[1]

Abstract

In road and off-road motorcycle accidents, neck injury often has a catastrophic outcome if not fatal. To protect motorcyclists' necks, a number of neck–braces are available on the market. The level of protection from these systems is not well reported because of the absence of an accepted neck-loading evaluation standard. The present study proposed a numerical evaluation for the neck–brace to protect the neck. Twenty-four impacts with and without the brace were simulated by changing impact angles and initial impact velocities. For each simulation, the vertical force, the extension moment, and the normalized neck injury criterion were calculated. Results showed that the risk of AIS 3 + neck injury was reduced by the neck–brace on average by 39 and 13% at 5.5 and 6.5 m/s, respectively, when the normalized neck injury criterion was considered; however, for impact velocities, above 6.5 m/s, the neck–brace was not as efficient in reducing overall neck injury risk.

Keywords Head–neck finite-element model · Neck–brace · Neck injury risk · Motorcycle

1 Introduction

Motorcyclists are considered as vulnerable road users, similar to pedestrians or cyclists, and account for up to 70% of road deaths in some Asian countries [1] and 18.3% in Europe [2]. Apart from the associated economic cost (medical care, etc.), traumatic injuries have a dramatic social impact, since they frequently lead to after-effect illness for the casualties. From an accidental point of view, Ooi et al. [3] have collected and analyzed 76 cases of motorcycle crashes, where cervical spine injury was reported. More than 50% of the cases presented an Abbreviated Injury Scale 3 + (AIS) for the cervical spine and 49% of the motorcyclists with an injury to the cervical spine additionally suffered head injury. A method to limit such consequences may be to develop protective systems which can transfer energy away from the neck during an impact event.

In the last few decades, several neck Finite-Element (FE) models were developed with varying levels of detail in terms of geometry, validation, and mechanical properties. Three model types can generally be defined. The first predicts the head trajectory [3–7], the second was developed to study whiplash injury [8–11], and finally, FE models were developed for axial loading dedicated to roll-over crashes or in the aeronautical domain [12–16]. Besides the model developed by White et al. [16], the other FE models were not able to extract forces and moments at the Occipital Condyle (OC), but only at T1 which is largely insufficient to extract meaningful neck injury criteria.

From an experimental point of view, under vertex loading, Mc Elhaney et al. [17] tested 14 surrogate cervical spine specimens with compressive impact velocity increasing until rupture (between 0.5 and 0.92 m/s). The average age of the cervical spine surrogates tested was 55 years. A large disparity between injury and loading location was observed. Only one sample presented an injury under compression–extension (intervertebral disc and ligaments at C4). The remainder of samples was injured under pure compression or compression–flexion. Under pure compressive loading in the upper cervical spine (C1–C2), two lesions at C1 and seven at C2 were identified. In the mid-to-lower cervical spine (C3–C7), five lesions in pure compression with associated vertebral body fracture were observed. The primary location of fractures was C5, C4, C3, and C6. Moreover, it was observed that

✉ Frank Meyer
ciaofok@gmail.com

1 Strasbourg University - ICUBE UMR 7357 CNRS, Strasbourg, France

Published online: 04 April 2018

injury occurring at lower cervical vertebral levels needed less energy than at the upper cervical spine (respectively, 5010 and 4060 N against 6040 and 6840 N). In 1984, Alem et al. [18] provided results on five Post-Mortem Human Subjects (PMHS) drop tests at various vertex loadings (7–11 m/s, impactor mass = 10 kg) and boundary conditions. Authors presented forces and moments at the C0 for five tests with no occurrence of injury. Nightingale et al. [19] proposed a study on the influence of boundary conditions on the injury mechanism in axial compression and compression–flexion. A total of 18 cervical spine surrogates were tested under three conditions at three velocities 0.045, 0.02, and 0.01 m/s. In addition, Nightingale et al. [20] performed 22 tests on the cervical spine in axial compression with differing impact conditions. Test subjects were positioned head down with an added mass of 16 kg fixed at T1 and dropped at a velocity of 3.2 m/s. Two impact surfaces were used, one rigid with a low coefficient of friction and one padded. Moreover, for each surface, three different impact orientations were created (− 15°, 0°, + 15°) to simulate posterior, vertex, and anterior head impacts. It was found that injury occurred more frequently with the padded surface than the rigid surface. Injuries recorded were of differing type and at several cervical levels. Cervical rupture forces were found to be significantly lower for female than male surrogates. For each test, resultant force, axial force, shear force, and impact force were reported. Pintar et al. [21] reported 20 tests under pure compression with a constant head impact velocity (between 2 and 8 m/s). The cervical spines were constrained to suppress their natural lordotic shape. Lesions obtained were compression related fractures, posterior ligament damage on compression–flexion, and anterior ligament rupture on compression–extension. No significant difference was noticed in terms of rupture force for these three injury mechanisms. Pintar et al. [22] reported on 13 tests conducted under axial compression with a constant impact velocity (between 2 and 5 m/s). To analyze flexion during impact, the authors defined three separate spinal columns between C0 and T1, namely, anterior, central, and posterior columns. Injury could then be classified in two stages, namely, a progressive flexion of the entire cervical spinal unit leading to a complex ligamentous system rupture in the posterior column. Second, the cervical spine may produce compression followed by a level-specific extension (which may lead to fracture or subluxation of the articular processes).

In parallel with the characterization of neck injury mechanisms, injury criteria have been developed to estimate the clinical potential for a specific injury risk scenario. In frontal impact analyses, the common criterion used is the Neck Injury Criterion (NIC) used at EuroNCAP, JNCAP, CNCAP, and LatinNCAP. It is defined by corridors in terms of extension moment, tension/compression, and shear force measured at OC. The normalized neck injury criterion (N_{ij}) used by US-NCAP, IIHS, and FMVSS combine normalized axial load and sagittal bending moments from the upper neck load cell. This criterion is not limited to frontal impacts and can be utilized for the evaluation of four different impact types, i.e., tension/extension, tension/flexion, compression/extension, and compression/flexion, and is, therefore, frequently used for motorcycle impact analysis. In rear impact, the most common criterion used is the NIC value developed by Boström et al. [23] on the difference in kinematics between the head and the first thoracic vertebra. The N_{km} criterion developed by Schmitt et al. [24] proposes the same formulation as the $N_{ij,}$ but other critical forces and moments are included and it is mostly limited to hyperextension analysis. Finally, the lower neck load (LNL) criterion is less frequently used and is calculated by a combination of forces and moments recorded or calculated in the three primary directions. Finally, there is no criterion defined for roll-over situations or lateral impact in the automotive domain. For motorcyclists, N_{ij} is the most commonly used criterion, since that all combinations of mechanistic impact scenarios, i.e., flexion/extension; tension/compression can be analyzed.

A first attempt towards a numerical evaluation of motorcyclist's neck loading under accidental conditions was reported in [25], where a helmet model was coupled to HUMOS model, without any neck–brace. This study computed the relative vertebra motions under different loading conditions, but did not express any injury risk.

In the present paper, a neck–brace and a validated helmet model [26] were coupled to an existing Head–Neck Finite-Element Model (FEM) [27] with a torso FEM coming from the THUMS V3 model. The previous validations of the Head–Neck FEM were only based on head kinematics (linear and angular accelerations, displacement, and head rotations) and done in terms of frequency response [27] under all directions. Because the present study is interested in vertex impacts, the Head–Neck FE Model needed to be validated in terms of neck forces under this type of loadings, as reported by Nightingale experiments [20]. To evaluate the ability of the neck–brace to protect the neck, 24 impacts have been simulated with and without neck–brace, by changing the impact angles and impact speeds. For each simulation, the vertical force, the extension moment, and the N_{ij} criterion were computed. According to the NHTSA N_{ij} "S-Curve", this parameter permitted us to assess the percentage of neck injury risk for AIS 3 + injury and to highlight the efficiency of the neck–brace.

2 Materials and methods

2.1 Strasbourg University Finite-Element Head Neck Model (SUFEHN-Model)

The Strasbourg University Finite-Element Head–Neck Model (SUFEHN-Model) is composed by 491 beam elements 110,786 shell elements and 10786 solid elements and consists of the coupling of the Strasbourg University Finite-Element Head Model (SUFEHM) improved by Deck et al. [28] with the Neck FEM developed by Meyer et al. [11]. Concerning the SUFEHM, the main anatomical features were taken into account, i.e., brain, brainstem, cerebellum, the two brain membranes (falx and tentorium) as well as skull, face, and cerebro-spinal fluid. The neck, cervical vertebrae, intervertebral discs, ligaments, and muscles were modeled. An illustration of the SUFEHN-Model is presented in Fig. 1a.

Previously [11], the SUFEHN-Model was validated against the N.B.D.L tests for frontal, lateral, and oblique responses [29, 30] and in the frequency domain to reproduce the well-known "S-Shape" [31]. This frequency-domain validation was extended for the frontal and transverse planes against Gunzel et al.'s [32] experiments. Based on the reconstruction of 86 real-world rear impact cases, the SUFEHN-Model proposed an injury criterion based on the shear displacement at each level of the cervical spine [27].

To investigate neck injury risk under vertex loading, a neck FEM must be able to extract forces and moments at each level of the cervical spine, including OC and T1. This is due to all experimental characterizations under vertex impacts being expressed in terms of force and moment parameters [19, 20]. In the previous studies, the SUFEHN-Model [11, 27] was not able to extract level-specific forces and moments. To deal with this problem, the same method developed and proposed by White et al. [16] was applied to the SUFEHN-Model. Therefore, for this study, the cervical vertebrae initially proposed with shell elements were remeshed with solid elements (increasing the number of solid elements to 55,951) and cross sections at each cervical level have been defined with its own local frame. An illustration of coordinate systems associated at each cervical vertebrae as well as cross-sectional location defined with the updated SUFEHN-Model is presented in Fig. 1b.

For the purpose of validating the presented updated neck model, Nightingale et al.'s [20] experiments have been used. The authors' conducted tests using a drop track system to produce impact velocities of the order of 3.2 m/s. Multi-axis transducers recorded head impact forces, head accelerations as well as forces at T1. An experimental apparatus was designed to model cervical spine injury resulting from vertical head impact with a torso. A steel carriage was mounted to a drop track using two linear bearing sliders and was weighted to simulate an effective torso mass of 16 kg. In this study, two types of impact surfaces were utilized, i.e., rigid and padded. The rigid impacts resulted in significantly larger peak head forces (4024 $\pm$ 1335 N) and shorter impulse durations than the padded impacts. However, the resultant neck forces at the time of neck injury were not significantly different.

For the validation of the updated SUFEHN-Model, experiments depicting the rigid surface were used. The

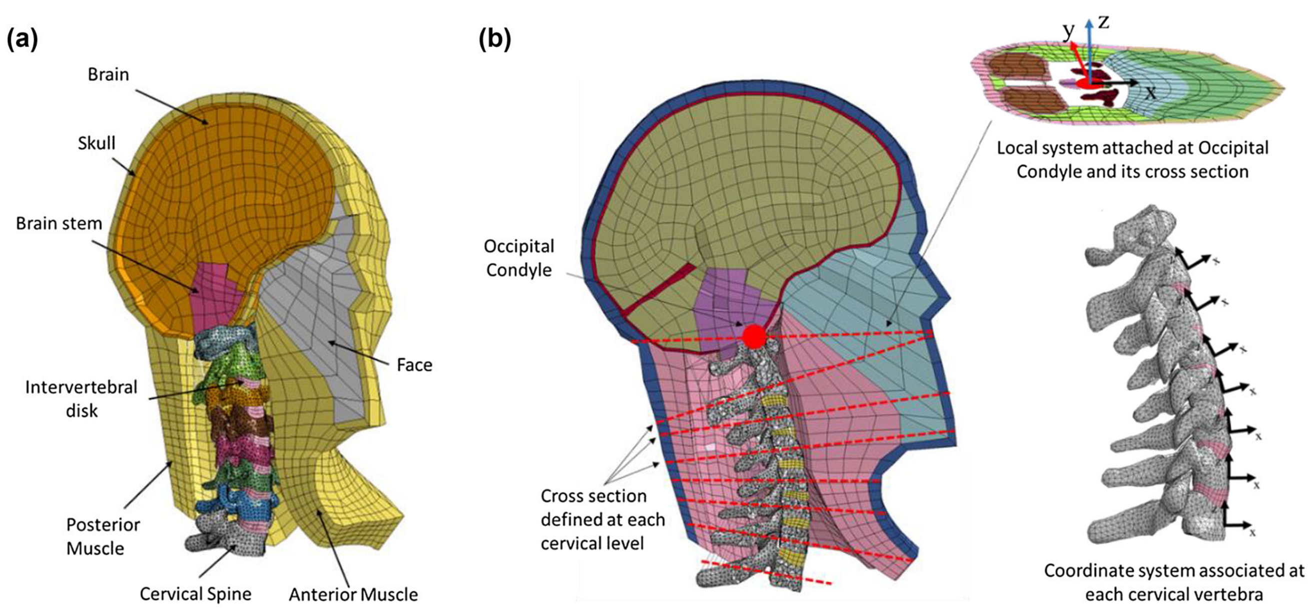

Fig. 1 **a** Cross section of the Strasbourg University FE Head Neck model (SUFEHN-Model), **b** definition of the section cuts and the coordinate systems associated with the cervical vertebrae and at the OC

padded surface tests induced further complexity in terms of mechanical characterization and friction response. Muscular tissue was removed from the model while keeping all the ligamentous structures intact in accordance with the experimental test setup. In addition, an added mass of 16 kg was applied to T1 level. Three impact angles were defined, namely, − 15°, (posterior head impact), 0° (pure vertex head impact), and + 15° (anterior head impact) as done in the experiments. Multi-axis model measurements were used to fully quantify the forces acting on the head and neck during the impact event and were compared to experimental records to validate the updated SUFEHN-Model against vertex loading.

2.2 Helmet FEM

For the present study, a motorcycle helmet finite-element model developed by Tinard et al. [33] with a composite shell and Expanded Polystyrene (EPS) foam was considered (Fig. 2a). The geometry of the helmet outer shell and EPS inner layer was provided by the helmet's manufacturer. The outer shell was meshed with 7542 shell elements with a thickness varying from 2.7 to 3.5 mm according to the considered helmet area. The EPS foam was meshed with 13,145 brick elements and 398 tetrahedron elements with thickness ranging from 13 mm for the chin up to 46 mm for the vertex. The entire model thus consisted of 21,085 elements. Mechanical properties implemented for each part of the helmet model were described in [26, 33]. The helmet model was coupled to a headform model to validate it under the standard impact conditions. The validation of the helmet model was performed using experimental data provided by the manufacturer including linear acceleration of the headform recorded during the same impact [33]. Impacts were carried out for different impact locations using kerbstone and flat anvils as required by ECE 22.05. Experimental versus numerical headform response in terms of head acceleration for a vertex impact against a flat anvil at 7.5 m/s is presented in Fig. 2b and the full validation is available in [33].

2.3 Coupling of updated SUFEHN-model, torso, helmet, and neck–brace FEMs

The neck–brace is designed to reduce the likelihood or severity of a neck injury during a motorcycle crash. It is used in conjunction with a full face helmet. It is designed as a semi-rigid platform sitting around the wearer's neck and aids in transmitting energy from the helmet to the body during compressive loading combined with flexion/extension or lateral rotation, bypassing the neck through the creation of an alternative load path onto other (safe) structures of the upper body.

For the present study, the CAD model came from an existing neck–brace. The model was created with a smooth surface and meshed with shell elements with an average element size of 5 mm. In this study, the neck–brace was assumed to be rigid. This simplification was motivated by the fact that no deformation and no damage to the neck–brace were observed during experimental tests. This was also the design intent of the manufacturer to create a theoretically more efficient path for load transfer away from the neck into the upper torso supports during helmet and neck–brace interaction for a typical neck injury mechanism. The neck–brace interacts with the torso during a vertex impact; therefore, the external surface of the torso THUMS V3 was coupled to the SUFEHN-Model. The connection was modeled with a spring element, adjusting for the decoupling between the shoulder and the head–neck system. To simulate the mass of the torso, a node with a mass of 65 kg was applied at the center of gravity of the torso. A complete overview of the FE model is given in Fig. 3.

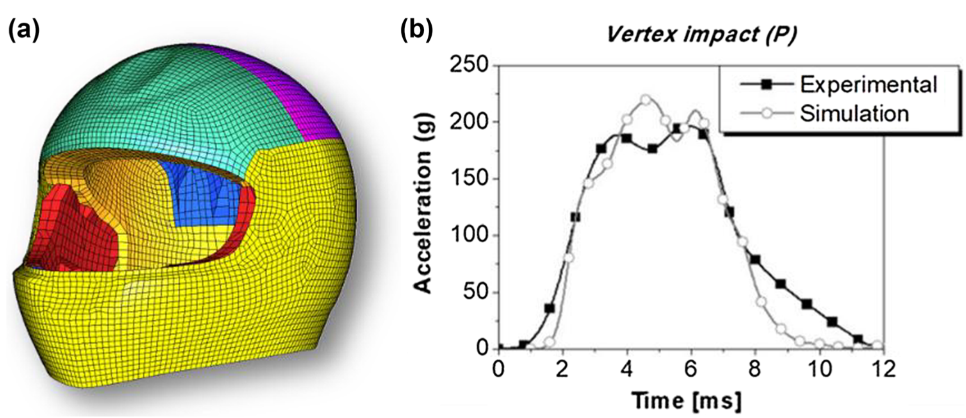

Fig. 2 **a** Illustration of the helmet finite-element model [33] and **b** its validation against a flat anvil at vertex area in accordance with ECE R022 standard [33]

Fig. 3 Description of the FEM composed by the updated SUFEHN-Model, Torso, Helmet, and Neck–Brace FEMs

2.4 Numerical impacts for neck–brace evaluation

To evaluate the level of protection offered by the neck–brace during a vertex impact, the full helmet–head–neck–torso–neck–brace system was dropped at four initial velocities, i.e., 5.5, 6.5, 7.5, and 8.5 m/s at three different angles. Angles were defined by taking into account head center of gravity and impact point axis and *X*-axis, as illustrated in Fig. 4; angles of 80°, 90°, and 100° relative to the *X*-axis for each velocity were used (0° corresponds to a frontal impact and 90° corresponds to a pure vertex impact).

The initial velocity was imposed to the whole model except the impact surface which was fixed and defined as a rigid body. To evaluate the efficacy of the neck–brace, baseline (no neck–brace) simulations were conducted for each impact speed and orientation, resulting in a total of 24 impact configurations (12 with neck–brace and 12 without neck–brace system).

For each of the simulated impacts (with and without neck–brace), results were computed in terms of axial forces (F_z) and flexion–extension moment (M_{OCY}) at the OC point (illustrated in Fig. 1b) to calculate the normalized neck injury criterion, N_{ij}, as the sum of the normalized loads and moments:

$$N_{ij} = \frac{F_z}{F_{zc}} + \frac{M_{Ocy}}{M_{yc}} \tag{1}$$

where F_z is the axial force, M_{Ocy} is the moment around OC, F_{zc} = 6160 N, and M_{yc} = 310 N.m (intercept values) according to NHTSA [34].

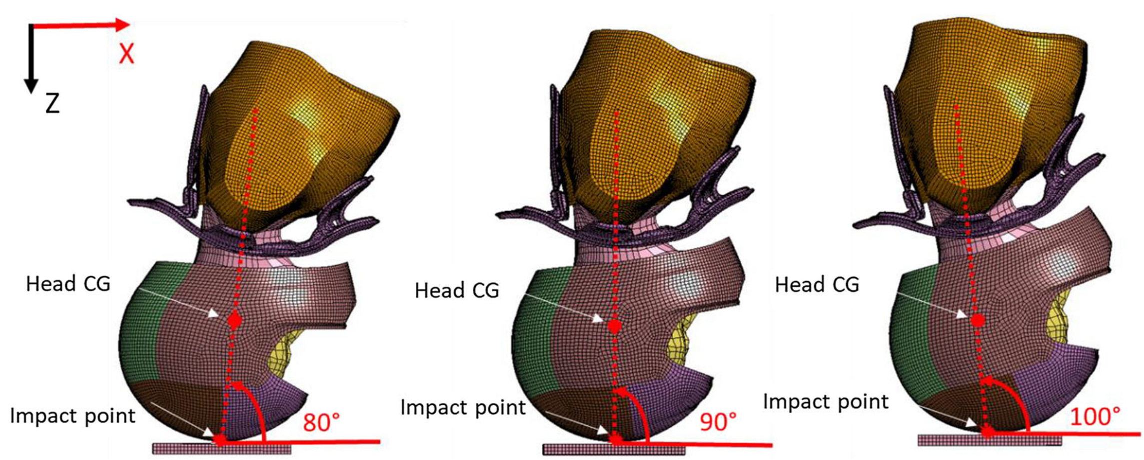

Fig. 4 Impact angle configurations (80°, 90°, and 100° relative to *X*-axis) used to evaluate the level of protection offered by the neck–brace during vertex impacts

3 Results

3.1 Validation of the updated SUFEHN-model

For the kinematic boundary condition, T1 was constrained in all directions besides vertical translation. An initial velocity of 3.2 m/s was applied to the entire head–neck system. The impact surface was defined as a rigid body with a friction coefficient of 0.2 [20]. Time simulation was over a 50 ms duration; the impact force between the impact surface and the head was computed with a contact time history. Concerning the neck response, the forces (axial, shear, and resultant) were calculated in the cross section of the first thoracic vertebrae.

Figure 5 illustrates the numerical validation for the vertex impact simulation in terms of head impact force and neck forces at T1 (curves were filtered with a 1000 Hz cut-off frequency, Post et al. [35]).

3.2 Evaluation of the neck–brace system

All results are presented in Tables 1 and 2 for simulations with and without neck–brace, respectively, and superimposed onto the neck injury risk curve proposed by NHTSA [34] (Fig. 6b). A representation of the results in the moment/force plane is also presented in Fig. 6a.

Most of the scenarios simulated induced a high level of probability for an AIS 3 + neck injury due to the severe nature of the impacts. For the baseline simulations (without the neck–brace), all the simulations presented a risk of AIS 3 + higher than 90%. For simulations with the neck–brace and for the lowest velocity (v = 5.5 m/s), the protective system reduced the risk by 47, 39, and 27%, respectively, for 80°, 90°, and 100° impact orientations. At 6.5 m/s, the neck–brace was less effective and reduced the risk by 21% at the 80° impact orientation. For impact velocities above 6.5 m/s, the neck–brace reduced loading, although the overall efficacy of cervical spine protection was limited.

To illustrate the typical behavior of the cervical spine in such an impact scenario, a simulation progression in terms of kinematics and mechanical response is shown in Figs. 7 and 8 for a 5.5 m/s and 100° impact event with and without the neck–brace, respectively. Axial forces and moments calculated at each cervical level, including OC are given in Fig. 9. After only 4 ms, the neck–brace interacted with the helmet and transferred load away from the neck, while without the neck–brace in position, the surface defined by the contact between the head and foam impact surface acted in compression, transferring load solely to the neck. For simulation without neck–brace system, after 20 ms of simulation, the EFS compression exceeded 80% and the stiffness of this part conducted to a “bottom off” of the head against the outer shell due to extreme loading.

Assessing the cervical spine axial compressive load, all level-specific forces with the neck–brace in position were found to be of the same order of magnitude (approx. 8000 N) and presented similar curve shapes, indicating a stabilizing/normalizing effect with the device (Fig. 9b). Without the neck–brace, forces were not equally distributed. Measured at T1, the maximum force was in the

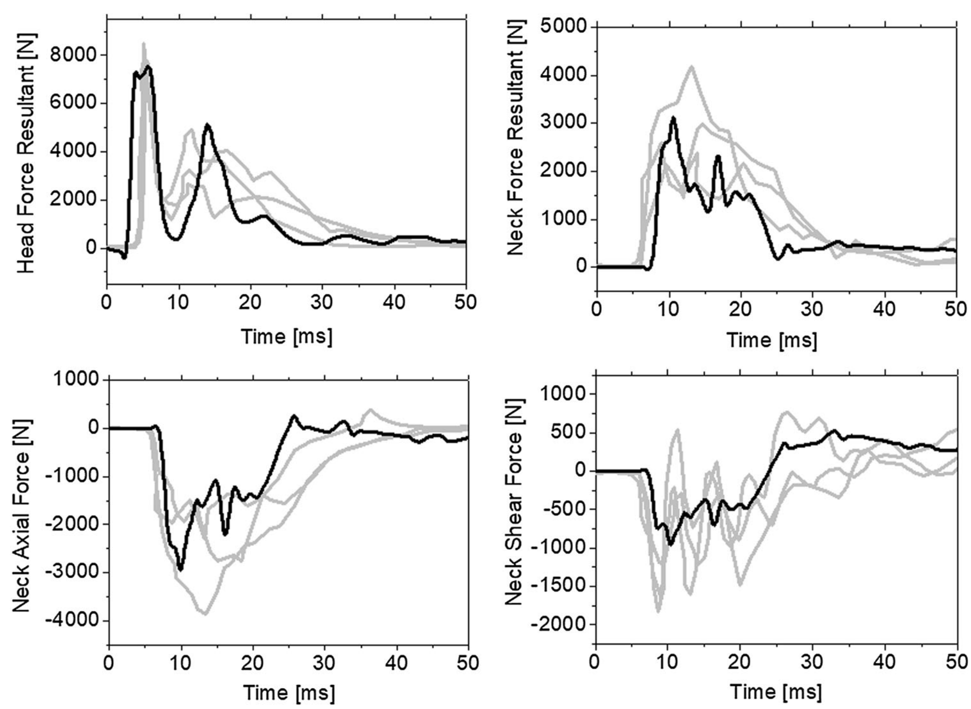

Fig. 5 Numerical validation of updated SUFEHN-Model (black curves) against Nightingale et al. [20] experiments (grey curves) for a vertex impact at 0° relative to *Z*-axis

Table 1 Compression force F_z, extension moment $M_{OCY,}$ and N_{ij} criterion in the three orientations for the four impact velocities with the neck–brace system

Impact speed (m/s)	80°			90°			100°		
	F_z [N]	M_{OCY} [Nm]	N_{ij} [−]	F_z [N]	M_{OCY} [Nm]	N_{ij} [−]	F_z [N]	M_{OCY} [Nm]	N_{ij} [−]
5.5	6027	196	1.6	7000	229	1.7	8177	204	2
6.5	7552	319	2.3	10,534	386	2.5	11,272	330	2.9
7.5	10,973	467	3.3	13,774	545	3.5	14,302	412	3.7
8.5	13,910	687	4.5	16,295	677	4.4	16,591	534	4.4

Table 2 Compression force F_z, extension moment $M_{OCY,}$ and N_{ij} criterion in the three orientations for the four impact velocities without the neck–brace system

Impact speed (m/s)	80°			90°			100°		
	F_z [N]	M_{OCY} [Nm]	N_{ij} [−]	F_z [N]	M_{OCY} [Nm]	N_{ij} [−]	F_z [N]	M_{OCY} [Nm]	N_{ij} [−]
5.5	11,744	382	3.1	13,021	333	3	13,358	246	3
6.5	13,358	483	3.7	15,180	421	3.5	14,689	486	4
7.5	15,289	515	4.1	16,379	501	4.1	15,521	336	3.6
8.5	16,230	611	4.6	18,115	548	4.4	15,974	315	3.6

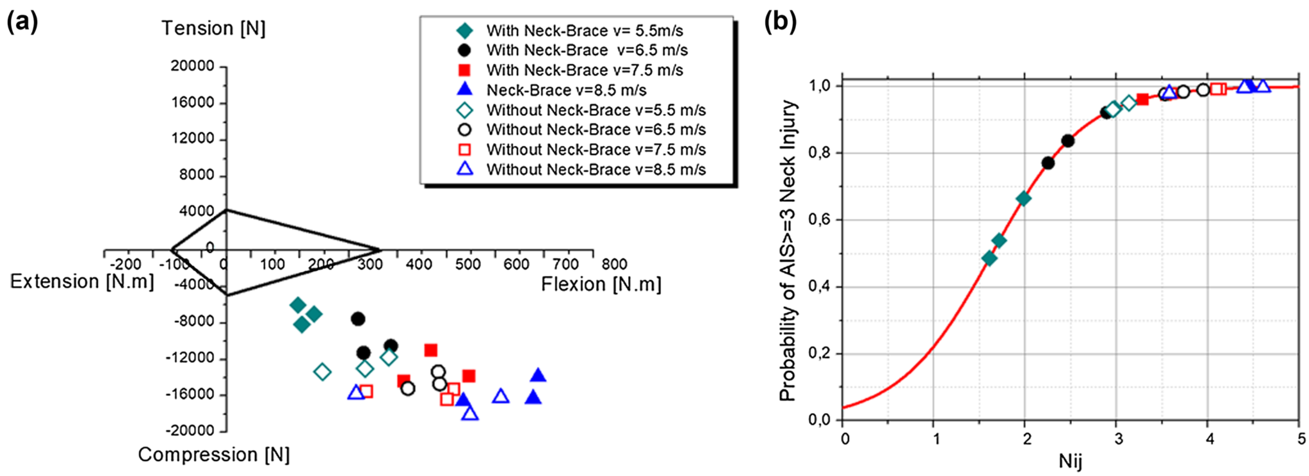

Fig. 6 Representation of the N_{ij} results computed (symbols) as compared to the NHTSA requirement (black line) and **b** neck injury risk of AIS 3 + related to N_{ij} for the 24 simulations computed with SUFEHN-Model

region of 12,000 N and at C6, 9200 N (Fig. 9a). Bending moments calculated with the neck–brace (Fig. 9b) showed the cervical spine in extension (due to its initial lordotic curve) and the head in flexion. For the baseline analysis, C6–T1 presented bending moments in extension with C1–C5 oscillating between level-specific flexion and extension. As in the first case, the head and OC were in flexion throughout the entire simulation. In all cases analyzed, both bending moments and axial loading were lower with the neck–brace as compared to baseline simulations.

4 Discussion

The present study suggested a neck–brace evaluation method based on numerical simulation. Even if validated neck models are known to be more biofidelic than dummies, there are still a number of critical issues. First, the neck model validation under vertical loading is a critical issue as only a limited number of experimental data recorded on PMHS are available in the literature. Therefore, the validation of the SUFEHN-Model is mainly restricted to these data as for the validation suggested by

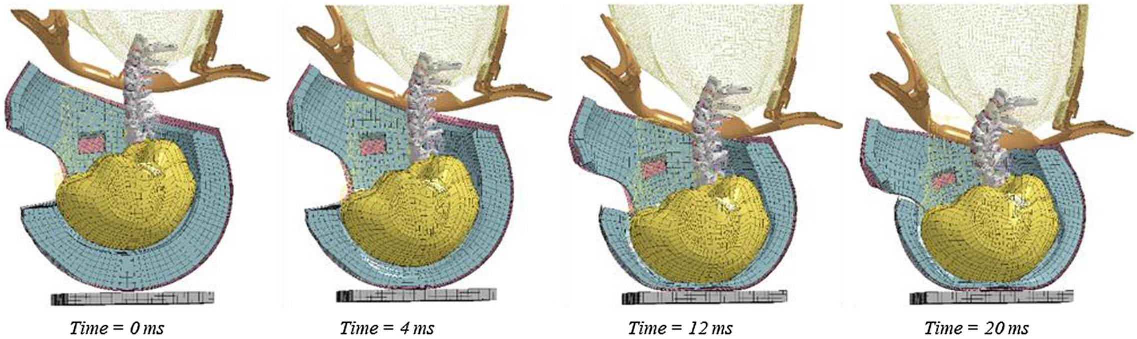

Fig. 7 Response of the SUFEHN-Model with the neck–brace system under a vertex impact at 5.5 m/s and at 100°

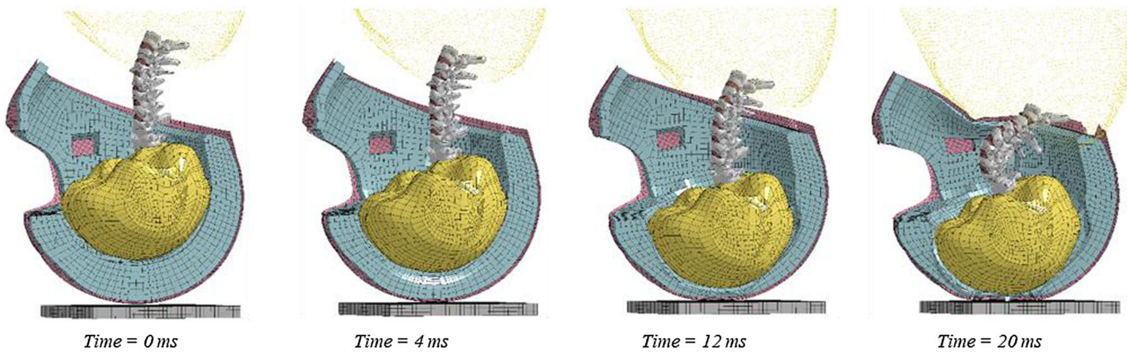

Fig. 8 Response of the SUFEHN-Model without the neck–brace system under a vertex impact at 5.5 m/s and at 100°

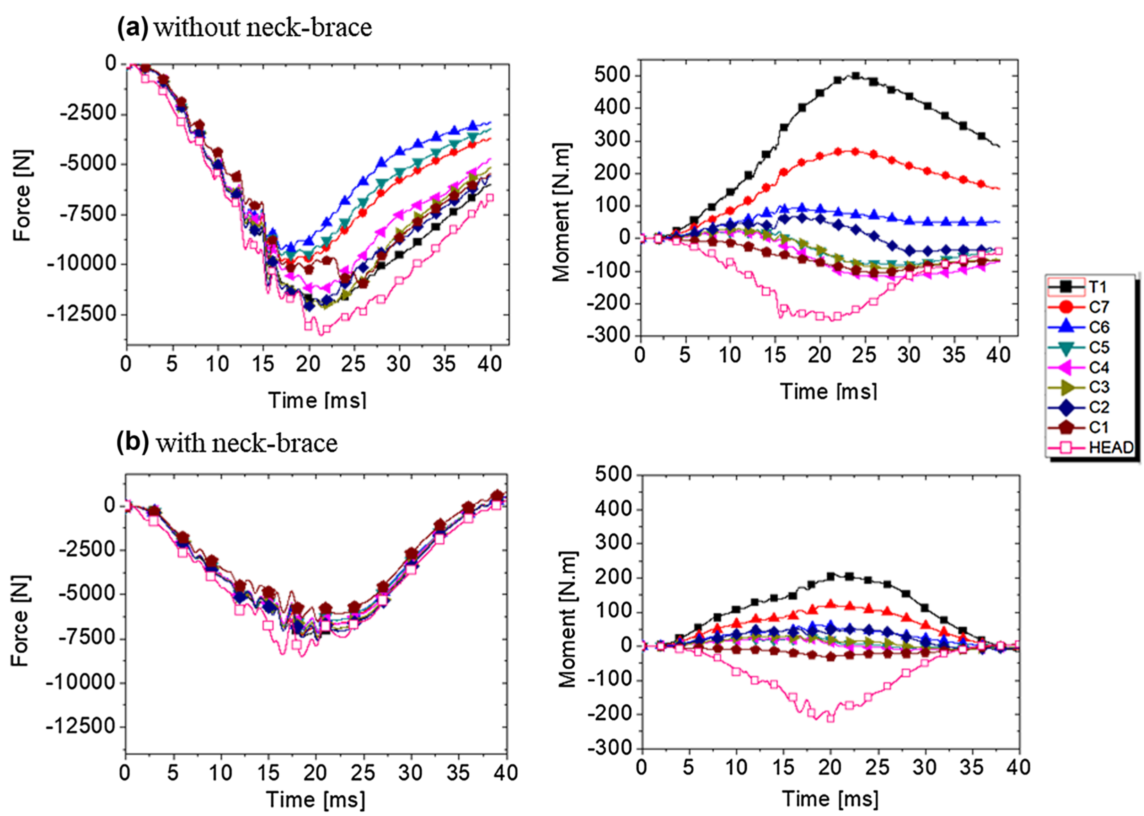

Fig. 9 Forces–time and moments–time traces calculated at different levels of the cervical spine with the SUFEHN-Model for a vertex impact at 5.5 m/s, 100°, without and with neck–brace

several models [12–15]. The proposed neck model further enables it to compute forces and moments between each vertebra and at C0/C1 level. Therefore, it was possible to assess the neck injury risk based on N_{ij} data, even if this injury criterion has been established for dummies and corresponds to severe AIS3 neck injury. There is clearly a need for a better understanding of moderate or acceptable neck injury and relevant injury criteria.

The proposed motorcyclist impact simulation involved a simplified coupling between the neck–brace and the shoulder represented by a simple spring. As this contact and the shoulder position play a critical role in force transfer, a further investigation should consider a parametric study in these effects. Moreover, an additional limit of the present study concerns the neck–brace FEM mechanical assumption done as a rigid body. Finally, the most extreme body angles were considered. Further development should extend the impact conditions to a wider range of body angles.

5 Conclusion

The updated SUFEHN-Model was validated and able to extract forces and moments at each cervical level and at the occipital condyle. A validation in vertex loading against tests in three different impact locations was performed. Torso, neck–brace, and motorcyclist helmet were coupled to the SUFEHN-Model to simulate 24 different impact conditions and gauge the efficacy of the neck–brace. The risk of injury, estimated against the N_{ij} criterion, was reduced by the neck–brace for velocities lower than 5.5 m/s for all impact orientations; however, for impact velocities above 6.5 m/s, the neck–brace was not as efficient in reducing overall neck injury risk, even though neck loading (forces and moments) was significantly reduced in most cases. It is expected that the present study contributes to the definition of neck–brace evaluation methods and is a step towards neck protection system standards.

References

1. Hsu TP, Ahmad FMS, Nguyen XD (2008) A comparison study on motorcycle traffic development in some Asian countries—case of Taiwan. Eastern Asia Society for Transportation Study, Malaysia and Vietnam, pp 1–53
2. European Road Safety Observatory (2014) Annual Statistical Report
3. Ooi SS, Wong SS, Yeap JS, Umar R (2011) Relationship between cervical spine injury and helmet use in motorcycle road crashes. Asia Pac J Public Health 23(4):608–619
4. De Jager M, Sauren A, Thunnissen J, Wismans J (1996) A global and a detailed mathematical model for head-neck dynamics. In: Proceeding of stapp car crash conference, paper 962430, pp 269–281
5. Astori P, Raballo M, (1998) Multi-body numerical model of the human neck. In: Proceeding international journal of crash conference, pp 63—72
6. Nitsche S, Krabbel G, Appel H, Haug E (1996) Validation of a finite-element-model of the human neck. In: Proceeding of IRCOBI conference
7. Happee R, Hoofman M, Van den Kroonenberger AJ, Morsink P, Wisman J (1998) A mathematical human body model for frontal and rearward seated automotive impact loading. In: Proceeding of stapp car crash conference, pp 75–88 paper 983150
8. Kobayashi S, Kitagawa Y (2003) Development of a finite element model of the human neck for whiplash simulation. In: Proceedings of the 18th international technical conference on the enhanced safety of vehicles, Nagoya
9. Deng Y, Fu J (2002) Simulation and identification of the neck muscle activities during head and neck flexion whiplash. In: SAE technical paper 2002-01-0017
10. Ono K, Ejima S, Sato F, Pramudita JA, Kaneoka K, Ujihashi S (2009) Evaluation criteria for the reduction of minor neck injuries during rear-end impacts based on human volunteer experiments and accident reconstruction using human FE model simulations. In: Proceeding of IRCOBI conference 2009, York
11. Meyer F, Bourdet N, Deck C, Willinger R, Raul JS (2004) Human neck finite element model development and validation against original experimental data. In: Proceeding of stapp car crash conference paper 2004-22-0008, pp 177–206
12. Camacho DL, Nightingale RW, Robinette JJ, Vanguri SK, Coates DJ, Myers BS (1997) Experimental flexibility measurements for the development of a computational head-neck model validated for near-vertex head impact. In: Proceeding of stapp car crash conference, paper 973345, pp 473–486
13. Halldin PH, Brolin K, Kleiven S, Holst HV, Jakobsson, L, Palmertz C (2000) Investigation of conditions that affect neck compression-flexion injuries using numerical techniques. In: Proceeding of stapp car crash conference, paper 2000-01-sc10
14. Yang K, Zhu F, Luan F, Zhao L, Begeman P (1998) Development of a finite element model of the human neck. In: Proceeding of stapp car crash conference, paper 983157, pp 195–205
15. Zhang QH, Teo EC, Wan HN (2005) Development and validation of a C0–C7 FE complex for biomechanical study. J Biomech Eng 127(5):729–735
16. White NA, Moreno DP, Gayzik FS, Stitzel JD (2015) Cross-sectional neck response of a total human body FE model during simulated frontal and side automobile impacts. Comput Methods Biomech Biomed Eng 18(3):293–315
17. McElhaney J, Paver J, McCrackin H, Maxwell M (1983) Cervical spine compression responses. In: Proceeding of stapp car crash conference, paper 831615, pp 163–177
18. Alem N, Nusholtz G, Melvin J (1984) Head and neck response to axial impacts. In: Proceeding of stapp car crash conference, paper 841667, pp 275–288
19. Nightingale R, Myers B, McElhaney J, Doherty B Richardson W (1991) The influence of end condition on human cervical spine injury mechanisms. In: Proceeding of stapp car crash conference, paper 912915, pp 391–399
20. Nightingale R, McElhaney J, Camacho L, Kleinberger M, Winkelstein B, Myers B (1997) The dynamic responses of the cervical spine: buckling, end condition, and tolerance in compressive impacts. In: Proceeding of stapp car crash conference, paper 973344, pp 451–471
21. Pintar F, Yoganandan N, Voo L, Cusick J, Maiman D, Sances A (1995) Dynamic characteristics of the human cervical spine. In: Proceeding of stapp car crash conference, paper 952722, pp 195–202

22. Pintar F, Voo L, Yoganandan N, Maiman D (1998) Mechanisms of hyperflexion cervical spine injury. In: Ircobi Conf, pp 249–260
23. Boström O, Svensson M, Aldman B, Hansson H, Håland Y, Lövsund P, Seeman T, Suneson A, Säljö A, Örtengren T (1996) A new neck injury criterion candidate based on injury findings in the cervical spinal ganglia after experimental neck extension trauma. In: Proceeding of IRCOBI conference, pp 123–136
24. Schmitt KU, Muser M, Walz F, Niederer P (2002) Nkm—a proposal for a neck protection criterion for low speed rearend impacts. Traffic Injury Prevent 3(2):117–126
25. Sun J, Rojas A, Bertrand P, Petit Y, Kraenzler R, Arnoux PJ (2012) Investigation of motorcyclist cervical spine trauma using HUMOS model. Traff Inj Prev 13(5):519–528
26. Tinard V, Deck C, Bourdet N, Willinger R (2011) Motorcyclist helmet composite outer shell characterization and modeling. Mater Des 32(5):3112–3119
27. Meyer F, Bourdet N, Gunzel K, Willinger R (2013) Development and validation of a coupled head-neck FEM—application to whiplash injury criteria investigation. Int J Crashworthiness 18(1):40–63
28. Deck C, Willinger R (2008) Improved head injury criteria based on head FE model. Int J Crashworthiness 13(6):667–679
29. Ewing C, Thomas D, Beeler G, Patrick L, Gillis D (1969) Living human dynamic response to -Gx impact acceleration II—Accelerations measured on the head and neck. In: Proceeding of stapp car crash conference paper 690817, pp 400–415
30. Ewing C, Thomas D, Lustick L, Muzzy W, Willems G, Majewski P (1977) Dynamic response of the human head and neck to +Gy impact acceleration. In: Proceeding of stapp car crash conference, paper 770928, pp 549–586
31. Bourdet N, Willinger R (2005) Human neck characterization under thoracic vibration—inter-individual and gender influence. In: Proceeding international conference on the biomechanics of impacts, pp 257–267
32. Gunzel K, Meyer F, Bourdet N, Willinger R (2009) Multi-directional modal analysis of the head-neck system and model evaluation. In: Proceeding of IRCOBI conference, pp 419–422
33. Tinard V, Deck C, Willinger R (2012) Modelling and validation of motorcyclist helmet with composite shell. Int J Crashworthiness 17(2):209–215
34. NHTSA (2008) Title 49 code of federal regulations (CFR) Part 571 section 208, occupant crash protection. Office of the Federal Register, National Archives and Records Administration, Washington, DC
35. Post A, Clark JM, Robertson DGE, Hoshizaki TB, Gilchrist MD (2017) The effect of acceleration signal processing for head impact numeric simulations. Sports Eng 20(2):111–119