AN ASSESSMENT OF THE CRASH PROTECTION AFFORDED BY NECK COLLARS FOR RACING CAR DRIVERS

D H Glaister, L P Neil and P J Waugh
Biodynamics Division,
Royal Air Force Institute of Aviation Medicine,
FARNBOROUGH, Hampshire, UK.

ABSTRACT

Four commercially available neck collars were evaluated in forward facing ($-G_x$) impacts using an instrumented Hybrid III dummy. Each was tested twice at $14ms^{-1}$ and $-25G_x$. A full-face helmet was worn and four control impacts were conducted without a collar. Head accelerations and HIC, and shear forces and moments about the upper and lower neck were recorded, and the impacts were monitored by high speed video (200 fps). Only one collar had any demonstrable effect, reducing the moment about the lateral head axis, M_y. It is concluded that proper specifications are needed to ensure the efficacy of neck collars for racing car drivers.

Neck collars, or neck supports, have become available on the vehicle accessories market with the aim, though not always stated, of supporting a helmet under high G-loadings and reducing injury to the neck in the event of a crash or rollover.

In order to investigate the efficacy of these crash aids, collars from six different manufacturers were made available for test through the Medical Committee of the RAC Motor Sports Association. Whole-body crash tests were conducted on the deceleration track at the RAF Institute of Aviation Medicine using an instrumented dummy, with particular reference to neck loads and head accelerations in forwards facing impacts.

METHODS

The test track uses a wheeled test vehicle which is accelerated to a pre-determined velocity using stretched rubber bungee cords and then coasts into a test area where it is subjected to controlled braking using a steel cable barrier and hydraulic pistons. The impact velocity was set at $14ms^{-1}$ and the hydraulic control orifices set to give a peak deceleration of 25G. Vehicle deceleration was monitored using a Philips PR 9367 accelerometer and the vehicle carried a 'generic' forwards-facing seat with 4-point harness. Typical vehicle deceleration traces are shown in figure 1.

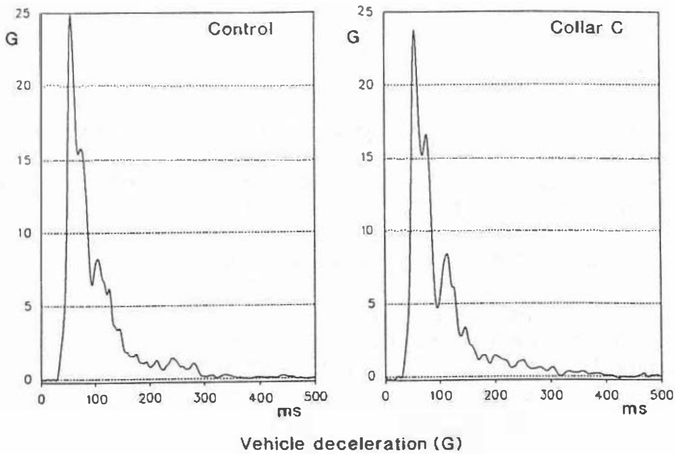


Fig.1. Vehicle deceleration recordings for run numbers 3628 (control) and 3629 (Collar C).

The dummy was a Hybrid III 50 percentile male (First Technology Safety Systems) fitted with the following load and acceleration transducers.

Table 1 - Anthropomorphic dummy instrumentation

Site	Transducer	Sensitive Axes
Head	Philips PR 9367/50	G_x, G_y, G_z
Upper Neck	Robert Denton Model 1794	$F_x, F_y, F_z, M_x, M_y, M_z$
Lower Neck	Robert Denton Model 1794	$F_x, F_y, F_z, M_x, M_y, M_z$

The signals from the vehicle accelerometer and the dummy transducers were conditioned using strain gauge amplifiers (Measurements Group Inc). The transducers were connected to the conditioning amplifiers by low loss screened flying leads. The conditioned signals were sampled using an analogue to digital sampling card (Metrabyte Das16) at 5000 samples per second. The sampled signals were filtered digitally using a frequency domain filter and a Blackman window with a cutoff frequency of 150 Hz.

The Denton convention was used to describe the force and moment axes. Since the dummy was symmetrical and subjected to a pure $-G_x$ (forwards deceleration) crash impulse, the recorded lateral forces (G_y, F_y) and the moments about the longitudinal (M_x) and vertical (M_z) axes were insignificant and are not considered further.

The dummy was positioned in the seat and the harness adjusted snugly before each run. It was fitted with a size 54/55 full-face thermoplastics helmet complying with BS 2495, and weighing 1.29kg. Due to the lack of 'soft tissue' surrounding the dummy neck, the test collars were centred so as to offer optimum support to the chin and helmet chinguard and then secured in position using adhesive tape. All impact sequences were monitored using a high speed video system (nac 200).

Four of the six collars were selected for test so as to cover the full range of perceived material properties in terms of stiffness and elasticity. Though differing in details, all were made up of a horseshoe shaped block of foam in a cloth cover, and could be secured round the neck by a touch and close fastening at the rear. In cross section the foams were rectangular, ranging from 50mm to 75mm wide by 45mm to 55mm deep, but varied widely in stiffness and resilience. For anonymity, the collars are referred to as A, B, C and D and twelve impact tests were conducted in the following sequence, the control condition being without any neck collar.

Table 2 - Details of test impacts

Run No	Impact Velocity ms^{-1}	Peak vehicle deceleration, G	Condition
3618	13.7	25.9	Control
3619	13.8	24.7	Collar A
3620	13.8	25.4	Collar A
3625	14.0	24.8	Control
3626	14.0	25.5	Collar B
3627	14.0	24.1	Collar B
3628	14.1	24.9	Control
3629	N/A	23.7	Collar C
3630	13.8	25.3	Collar C
3631	13.8	26.3	Control
3632	13.8	24.8	Collar D
3633	13.8	24.2	Collar D

RESULTS

As shown above, impact velocities and vehicle accelerations were consistent and showed no significant trend with run number, so that the four collars were treated consistently.

Impact data for run numbers 3628 and 3629 (control and collar C) are illustrated in figure 2, for head resultant acceleration, and in figure 3 for upper and lower neck M_y.

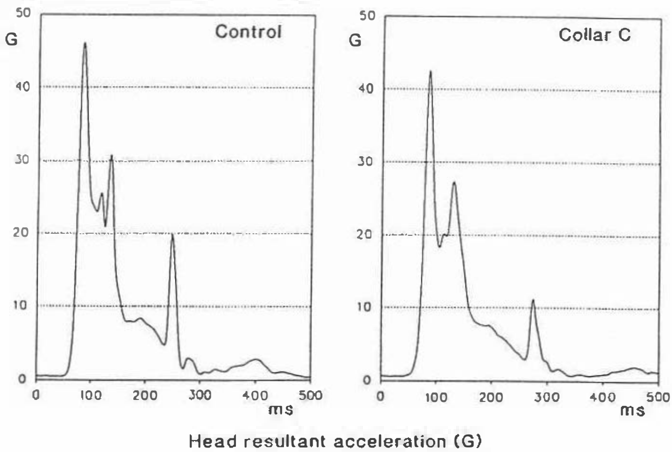


Fig.2. Head resultant accelerations for run numbers 3628 (control) and 3629 (Collar C).

The head resultant acceleration (Fig.2) shows considerable dynamic response, peaking at nearly two times the peak vehicle deceleration. Analysis of the individual acceleration vector recordings, together with the videos of the impacts, showed that the complex profile is made up from a $-G_z$ peak at 80ms as the head is thrown forwards, a $-G_x$ peak at 120ms as the chin contacts the dummy's chest, and a final $+G_x$ peak as the head rebounds on to the headrest. Neck collar C had little effect on these forces, though there was a tendency for the last peak to be attenuated, suggesting that some of the recoil energy had been absorbed.

Shear forces about the fore and aft axis of the neck (F_x) showed a similar response to the impact at both upper and lower sites, peaking at chin-chest contact with values of 300-400lb (1.3-1.8kN), and were unaffected by the presence of any of the neck collars.

Axial loading of the neck (F_z) showed a much spikier response, with the distraction force relieved during chin contact, particularly at the lower neck. Peak forces of 550-700lb (2.4-3.1kN) were essentially unaffected by any of the collars.

Neck torque about the lateral axis (M_y) peaked at chin-chest contact with a reversal as the head rebounded on to the headrest (Fig. 3). The moment is greater in the lower neck due to the longer lever arm available and collar C caused a small reduction, from 3,700 lb-in to 3,000 lb-in (from 420 N.m to 340 N.m) at this site. However, collar C did have a very dramatic effect in reducing M_y in the upper neck, from nearly 900 lb-in (100 N.m) to 350 lb-in(40 N.m).

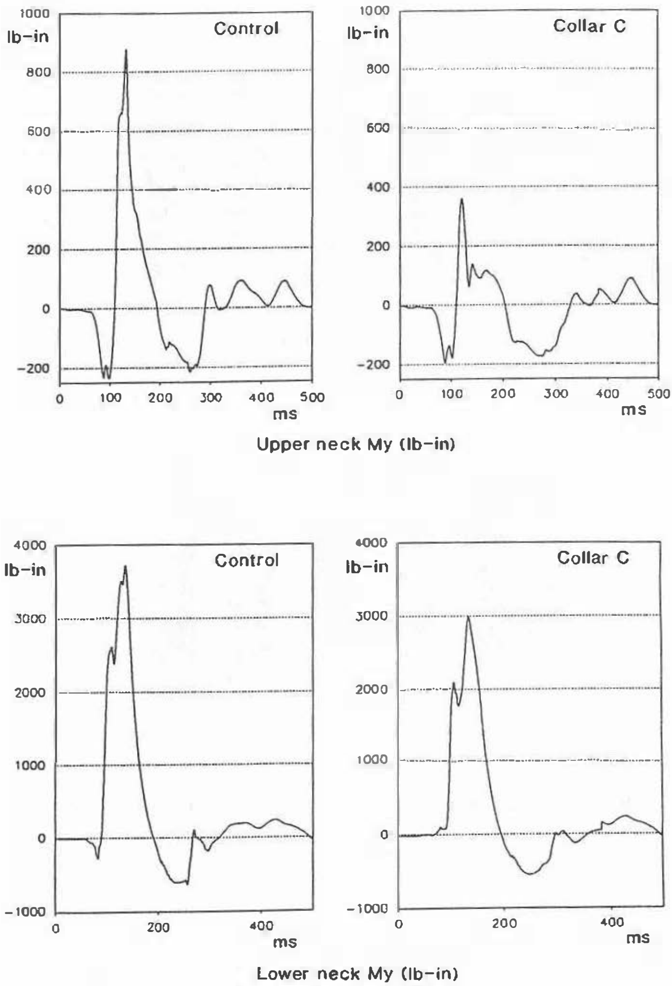
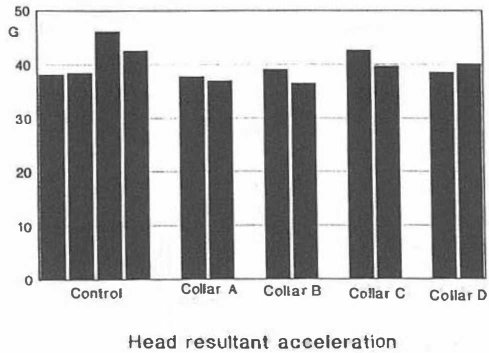


Fig.3. Upper and lower neck moments about the lateral axis (M_y) for runs 3628 (control) and 3629 (Collar C).

Relevant data from all the 12 control and neck collar runs are summarised in figures 4 and 5. Figure 4 (upper panel) shows head resultant accelerations, with all the neck collar values lying within the spread of values seen without neck support. Also shown in figure 4 (lower panel) are head injury criterion (HIC) values. There is a fair degree of scatter in the control values, but the HIC may have been slightly reduced by collar C, though the reduction was not statistically significant.



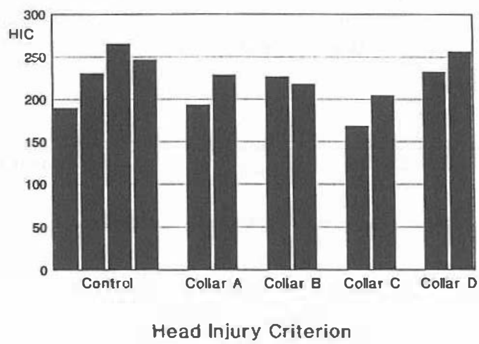
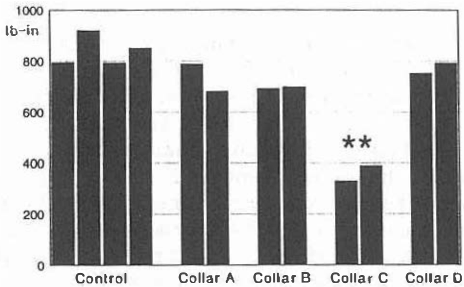


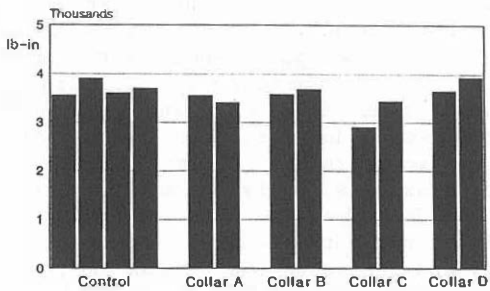
Fig.4. Peak head resultant accelerations (upper panel) and head injury criteria (lower panel) for all 12 runs.

Values for F_x and F_z showed no significant effect from any of the collars, but M_y (Fig.5) was greatly attenuated by collar C (P=<0.01) for the upper neck and possibly decreased by this collar for the lower neck. None of the other collars had any effect on M_y, lower neck, but there was a tendency for collar B to have caused a small reduction in upper neck M_y, though again, this finding was not statistically significant.

The video recordings were analysed to measure the maximum angular deflections of the helmet, and by inference, that of the dummy head about the Y axis, during chin contact and subsequent rebound, using the lower rim of the helmet as a postioning index. Average values obtained are given in table 3.



Upper Neck My



Lower Neck My

Fig.5 Peak neck moments about the lateral axis (M_y) for the upper and lower neck for all 12 runs

Table 3. Maximum helmet rotations

Condition	Forwards rotation	Rearwards rotation
Control	98^o	20^o
Collar A	92^o	17^o
Collar B	84.5^o	22^o
Collar C	79^o	14.5^o
Collar D	94^o	24.5^o

While data are too few for statistical analysis, collar C has clearly restricted both forwards and subsequent rearwards rotation, presumably because it had absorbed a significant amount of the impact energy. The other collars had little if any effect and collar D may even have increased the rebound somewhat.

DISCUSSION

Visual examination of the six collars offered for test showed that collar A contained a high density springy (and presumably closed cell) foam as did collar B. Collar C had the 'marshmallow' feel of a medium density polyurethane-silicone foam, while collar D was very soft, with the consistency of upholstery padding.

The two untested collars were also soft and springy and would be expected to perform like collar D. More detailed examination of the collars was not possible as they had all been loaned for the tests and had to be returned undamaged.

Of the eight measurements made which could be related to head or neck injury, only the moment about the lateral axis of the neck (M_y) showed any influence of collar wear, and then only for collar C. It may be noted that this collar was considered to contain a rate-dependent, energy absorbing foam. Collars with high density springy foams such as specimens A and B have the potential to exacerbate recoil forces, though this effect was not demonstrated. Collar B appeared to reduce head forwards flexion slightly and may have reduced M_y at the upper neck, though this reduction was not statistically significant. Collar D may even have stored impact energy and increased the rebound on recoil.

Concussion tolerance is generally accepted to be about 300G (Swearingen, 1971), or at a Head Injury Criterion value of 1000. Concussion would not, therefore, have been predicted for any of the impact tests, with or without the collars. This finding is as expected since it is generally accepted that, in motor vehicle crash tests, HIC values in excess of 1000 are only seen when the head makes direct contact with a rigid part of the vehicle structure.

In a forwards facing ($-G_x$) impact, neck injury could result from excessive flexion bending moment (M_y), excessive shearing force (F_x), or an excessive distraction force. (F_z). Mertz (1984) reviewed the then available test data and drew up guidelines for human tolerance, based upon the performance of the Hybrid III dummy. The reference value for neck flexion bending moment was 190 N.m (1,680 lb.in.). Figures 3 and 5 show that in the critical area of the upper neck, the recorded values were only some 50% of the reference tolerance level. Comparable data are not presently available for tolerances at the level of the lower neck. It may be predicted that with greater impact forces, the wearing of an effective neck collar (such as collar C) would have a significant effect in reducing the risk of neck injury.

The Mertz guidelines for axial neck distraction (tension) forces are time dependent, and the values recorded in the present tests of 400-700 lbs (1.7-3.1 kN), lasting for 10-12 ms, lie close to his tolerance boundary, but within his 'injury unlikely' zone. As would be expected from the impact geometry, the wearing of a neck collar has no effect on axial tension, nor could it influence neck shear forces. Guidelines for fore and aft shear are also time dependent and again, recorded values of 300-400 lb (1.3 to 1.8 kN), lasting for up to 60 ms, lie just below the boundary for significant neck injury.

The reported tests were restricted in that only $-G_x$ impacts were considered, and the impact velocity of $14ms^{-1}$ (50.4kph) and peak sled acceleration at 25G are less than could occur in many actual accidents. The design of the neck collar, a horseshoe closed behind the neck by a touch and close fastener, implies that when a full-face helmet is worn (as intended), the greatest volume of foam, and hence energy absorbing capacity, will be utilised in a forwards facing impact. Examination of the video recordings and the data of Table 3 show that the foams had effectively bottomed-out even under the modest $-G_x$ impact levels imposed. Thus, from the failure of three of the collars to demonstrate any benefit under these optimum test conditions, it may be concluded that no benefit could be exacted under more severe or realistically multiaxial conditions.

CONCLUSIONS

1. From the recorded neck data and published neck injury tolerance levels, the test impacts would not have produced significant neck injuries either with, or even without the neck collars, though neck shear forces were close to published tolerance limits. Neck shear forces were, however, unaffected by the wearing of any of the collars under test.

2. While excessive neck flexion bending moment can be a cause of serious injuries such as atlanto-occupital or C1-C2 separation (Nyquist and King, 1985), the forces recorded in these tests were well below published tolerance values for the upper neck. At higher impact loads, the wearing of collar C would be expected to reduce the risk of such injuries.

3. Only one of the four collars tested, and by inference, only one of the six types submitted for test, had any demonstrable effect in reducing neck forces in low level, forwards facing impacts. If neck collars are to be offered for sale and worn in motor sport activities, it is essential that appropriate specifications be drawn up so that their efficacy can be

assured, and potentially injurious devices avoided.

ACKNOWLEDGEMENTS

The authors are grateful to the Royal Automobile Club Motor Sports Association Medical Committee for provision of the test collars.

REFERENCES

Mertz H.J. Injury assessment values used to evaluate Hybrid III response measurements. General Motors Corp. NHTSA Docket Submission VSG 2284 Part III, Attachment 1, Enclosure 2. 1984

Nyquist G.W. and AI King. Review of Biomechanical Impact Response and Injury in the Automotive Environment. Chapter 2, Spine. Report PB87-182168 prepared for National Highway Traffic Safety Administration by Michigan University. Ann Arbor. March 1985.

Swearingen J.J. Tolerance of the human brain to concussion. Federal Aviation Administration Report FAS-AM-71-13. 1971.

Discussion of
AN ASSESSMENT OF THE CRASH PROTECTION AFFORDED BY NECK COLLARS FOR RACING CAR DRIVERS

Mats Y. Svensson
Department of Injury Prevention
Chalmers University of Technology
S-412 96 Göteborg, Sweden

ABSTRACT

The present study was carried out with the best crash test dummy currently available (Hybrid III) equipped with good instrumentation and the data were compared to established injury criteria. The bio-fidelity of the Hybrid III dummy neck is however questionable and this influences the validity of the results of this study. Results from recent studies, regarding sites and mechanisms of neck injuries implicate that neck collars, in order to have good effect, have to interfere with the forward translational and forward angular head-neck motion early on in the crash event.

The authors have chosen the Hybrid III dummy as a human surrogate and this was probably the best currently available crash test dummy for their purposes. The Hybrid III neck was designed to meet criteria regarding torque at the occipital joint as a function of head angular displacement relative to the torso. The head-neck kinematics of the Hybrid III dummy have, however, been shown to lack bio-fidelity in the frontal impact situation (Wismans and Spenny, 1984; Seemann et al., 1986; Deng, 1989; Mendis et al., 1989).

When comparing volunteer and Hybrid III head kinematics, Wismans and Spenny (1984) found that the maximum downward displacement of the head CG relative to the torso was about twice as high for a volunteer compared to the Hybrid III dummy. The maximum forward angular displacement of the head was also significantly higher for the volunteer compared to the Hybrid III. Mendis et al. (1989) presented a comparison of the relation between translational and angular head displacement, between the Hybrid III dummy and volunteer data. They found that the delay of the onset of forward angular head motion, relative to the onset of the forward translational motion, was more pronounced for the volunteers compared to the Hybrid III. The development of an improved dummy with better neck bio-fidelity is in progress (Eppinger et al., 1994) and will hopefully soon provide us with better means for assessment of neck injury risk in car collisions.

Considering the differences in head kinematics described above and the rather large difference in shape of the upper frontal part of the chest between dummy and human being, it is reasonable to assume that the results of the present study may have been significantly different with a more bio-fidelic dummy. The longer downward head displacement of a human being compared to a dummy would cause a larger compression and thus a larger effect of the neck collar than what the dummy tests indicate. On the other hand, the chest contour of the Hybrid III dummy probably gives a better support to the bottom of the collar compared to the human chest and this may cause an overestimation of the collar effectiveness.

According to the authors, the neck collars are intended to reduce the risk of neck injury in the event of a crash. The frontal impact situation was chosen for investigation in the present study and in this impact direction several types of neck injury could occur. The majority of car-accident induced neck injuries are of soft tissue type often denoted "neck sprain" or "whiplash injury" and are classified as AIS 1 (Hildingsson, 1991). A smaller part of the accidents result in higher AIS ratings involving fractures or obvious damage to discs, ligaments, facet joints etc. In rare cases these more severe injuries also lead to spinal cord injuries.

Neck injuries of AIS 1 occur in all impact directions but are most common in rear-end collisions. Several investigators, e.g. Deans et al.(1987) and Maimaris et al. (1988), have found the frontal-impact neck-injury incidence to be about 50% of that of rear-end impacts. The AIS 1 type of neck injuries cause a number of well documented symptoms but the sites of the injuries and the mechanisms behind them have not been established.

The neck-injury symptoms appear to be similar regardless of impact direction (Hildingsson, 1991) and this indicates that these neck injuries are of the same nature in both frontal and rear impacts. In the rear-end impact situation, findings of McConnell et al. (1993) show that these injuries are not caused by hyper-extension of the cervical spine. Findings of Svensson et al. (1993) from experimental studies on anaesthetised pigs indicate that injuries to the cervical spinal ganglia cause most of the typical symptoms of these AIS 1 neck injuries sustained in rear-end impacts and that the injuries may be caused by transient pressure gradients along the cervical intervertebral foramina. These pressure gradients were shown to occur already at the onset of the angular head-neck motion in both forward and rearward experimental "whiplash motion" (Svensson et al., 1993).

Frontal-impact dummy neck experiments were carried out by Deng (1989). He found that the highest linear and angular head accelerations and also the highest forward bending moment and virtually maximum shear at the occipital joint occurred very early during the impact event, at a time when the forward angular head motion was initiated. In the present study, the peak head acceleration similarly occurred early on before the neck collar began taking load. This explains the absence of collar effect on the peak head acceleration.

It appears that neck injuries, whether caused by pressure gradients or by shear loads and moments at the occipital joint, may well occur very early on in the crash event while only slight angular displacement of the head relative to the torso have occurred (Deng, 1989; Svensson, 1993). This means that a neck collar, in order to be effective, would have to interfere with the forward head-neck motion very early on and limit not only the forward angular head motion but also the forward translational head motion relative to the torso. This could not be achieved with the design concept of the collars tested in the present study and this is also pointed out by the authors. I fully agree with the authors' conclusion that it is essential that appropriate specifications become available so that the efficacy of this type of collars can be assessed.

REFERENCES

Deans, G.T.; Magalliard, J.N.; Kerr, M.; Rutherford, W.H.: Neck Sprain - a Major Cause of Disability Following Car Accidents. Injury, 18: 10-12, 1987

Deng, Y.-C.: Anthropomorphic Dummy Neck Modelling and Injury Considerations. Accid. Anal. & Prev., 21(1): 85-100; 1989

Eppinger, R.; Kleinberger, M.; Morgan, R.; Khaewpong, N.; Bandak, F.; Haffner, M.: Advanced Injury Criteria and Crash Evaluation Techniques. Paper no. 94-S1-O-11, Proc. Fourteenth Int. Conf. Enhanced Safety of Vehicles, Munich, May 23-26, NHTSA, Washington, USA, 1994

Hildingsson, C. (1991): Soft Tissue Injury of the Cervical Spine. Umeå University Medical Dissertations, New Series No 296, ISSN 0346-6612, ISBN 91-7174-546-7

Maimaris, C.; Barnes, M.R., Allen, M.J.: 'Whiplash injuries' of the neck: a retrospective study. Injury, 19: 393-396, 1988

McConnell, W. E.; Howard, R. P.; Guzman, H. M.; Bomar, J. B.; Raddin, J H.; Benedict, J. V.; Smith, L. H.; Hatsell, C. P.: Analysis of Human Test Subject Responses to Low Velocity Rear End Impacts. SAE paper no. 930889, SAE Inc. Warrendale, Philadelphia, USA, 1993

Mendis, K.; Stalnacker R.L.; Pritz, H.B. Multi Directional Neck Prototype. Proc. Twelfth Int. Techn. Conf. Experimental Safety Vehicles, US Dept. of Transp., NHTSA, USA, 645-649; 1989

Seemann, M.R.; Muzzy, W.H.; Lustick, L.S. Comparison of Human and Hybrid III Head and Neck Response. Proc. 30:th STAPP Car Crash Conf., SAE/P-86/189, 291-312; 1986

Svensson, M. Y.; Aldman, B.; Lövsund, P.; Hansson, H. A.; Seeman, T.; Suneson, A.; Örtengren, T.: Pressure Effects in the Spinal Canal during Whiplash Extension Motion - A Possible Cause of Injury to the Cervical Spinal Ganglia. Proc. 1993 Int. IRCOBI Conf. on the Biomechanics of Impacts, Eindhoven, The Netherlands, 189-200, 1993

Wismans, J. and Spenny, C.H. : Head-Neck Response in Frontal Flexion. SAE paper no. 841666, SAE/P-84/152, ISBN 0-89883-7111, 1984