Intracranial Injuries of Fatally Injured Pedestrians - A Contribution to the Establishment of Tolerance Limits

C. AUER, G. BEIER, E. SCHULLER

Institut für Rechtsmedizin der Universität München
(Director: Prof. Dr. med. W. Eisenmenger)

Abstract

The concern of this paper is the study of mechanical tolerance limits of intracranial injuries, i.e., epidural, subdural, subarachnoidal, coup and contre-coup contusions and brain lacerations, by the investigation of real world accidents.

Twenty-five fatal pedestrian accidents have been analysed with respect to pedestrian kinematics, head impact velocity, contact area, and mechanical loading as related to sustained head injuries. From this analysis, upper and lower tolerance limits are defined: Above a mean acceleration of 250 g subarachnoidal hematoma, above 300 g subdural hematoma, brain lacerations, and contre-coup contusions were *regularly* observed. Due to uncertainties in reconstruction and analysis, values marking the amount of loading below which these injuries were *never* observed are rather low and range between 20 and 80 g for long pulse durations (20-40 ms).

FURTHER EXAMINATION of the biomechanical tolerance limits of the human brain is required. Rather than carrying out the establishment of injury criteria the interest was in examining the incidence of particular injuries. If these are known, global injury criteria can be deduced. The knowledge of the limits of intracranial injuries is also of great importance for the analysis and reconstruction of actual accidents in accident research or for forensic purposes. The study is based on pedestrian accidents since these can be reconstructed and analysed in a comparatively good way. Our further interest was in the discrimination between vehicle and pavement impact in the origin of injuries and in the case of the former, which contact areas (hood, windshield, A-pillar) are most likely to cause certain injuries.

Materials and Methods

In the years 1990-1993 243 cases of fatally injured pedestrians involved in accidents with passenger cars were subject to forensic medical examination at the University of Munich Institute of Legal Medicine. From these, the course of the accident and the documentation in 25 cases allowed for the carrying out of a biomechanical and technical analysis with respect to the research topic.

Acquisition and Analysis Procedures

The accidents have been analysed in detail with regard to impact kinematics, mechanical loading to the head, impact duration, and resulting brain injuries.

Mean head acceleration, impact duration, and impact force were estimated from the contact area deformation, the head impact speed, and the basic acceleration histories. For rigid contact areas (e.g. windshield-frame, A-pillar, windshield) a triangular acceleration history and for soft contact areas (hood) a parabolic acceleration pulse were assumed.

Because of uncertainties of the input values which determine the mechanical loading only data ranges can be calculated. The parameters collision speed, head impact velocity, and dynamic contact area deformation were reconstructed and varied in such a manner that it is guaranteed that the actual loading was within these ranges. The results were verified by comparing these with results of cadaver and dummy tests published in literature [1], [2], [3], [14], [16], [23], [30], [31].

The steps in working up a pedestrian accident are as follows:

1. *Medical examination:* In every single case a complete forensic section was performed and the entire injury pattern was assessed. Injuries of the integument and soft tissue as well as bone fractures caused by contact with the vehicle were geometrically measured. The cranium was opened in the usual manner. The brain was cut in slices and the morphologically detectable injuries were located and diagnosed using the scheme of RYAN et al. [26]. Photographs of every injury were taken.
2. *Vehicle inspection:* In most cases the vehicle was available for inspection. All deformations were measured, especially the depth of the head contact dent. Photographs were taken. In a few cases car damage and deformations had to be taken from the photographic documentation of the police records.
3. *Tracks at the accident site:* From the police records the final positions of the pedestrian and the vehicle were taken, as well as skid marks, scratch marks, fragment patterns on the road, and items thrown away from the pedestrian. If necessary, the site was reinspected.
4. *Impact speed:* The collision speed was calculated using all obtainable data such as skid marks, throwing distance of the pedestrian, etc. If possible, the calculations were done using several methods [7], [8], [17].
5. *Impulse geometry and kinematics:* From the injury pattern, the vehicle damage, and the tracks at the accident site the relative orientation of the pedestrian in relation to the vehicle at the time of the first contact was assessed [4], [5], [6], [25], [27]. Using the injuries and the tracks on the vehicle and the road it could be reconstructed how the body and especially the head impacted the vehicle and the pavement, resp.
6. *Velocity of head vehicle impact:* The velocity of the head impacting the vehicle is dependent on the collision speed and the height of the upper leading edge (ULE) of the hood [23], [27]. JANSSEN et al. [15] and WALZ et al. [34] have shown that the angle between head impact direction and vehicle surface is almost perpendicular. Therefore a splitting into speed components is not necessary. The head impact velocity $v_{head\perp}$ was calculated from $v_{head} = c_{form} \cdot v_{coll}$ according to NIEDERER and SCHLUMPF [23]. The form-factor c_{form} determines the relation between collision speed v_k, upper leading edge of the hood (ULE), and head impact velocity. c_{form} ranges from 0.75 to 1.25. For high collision speed it reaches 1.0 which means that then the head impact velocity is rather independent of the front geometry of the vehicle.
7. *Head loadings during vehicle impact:* The dynamic deformation s_{dyn} was derived from the head contact dent on the car body, i.e., the permanent deformation due to head impact: The dynamic deformation for hood and windshield contacts was assumed to be 10 - 20% higher than the permanent deformation. Concerning contacts with the windshield frame, it was assumed that the dynamic deformation equals the permanent one. Using the calculated head impact velocity $v_{head\perp}$ the mean acceleration $\bar{a}$ and the impulse duration t_I were estimated: for triangular acceleration profiles (impact on A-pillar, windshield and windshield-frame) using

$$\bar{a} = \frac{v_{head\perp}^2}{2s_{dyn}} \quad ; \quad t_1 = \frac{2s_{dyn}}{v_{head\perp}} \qquad (1),$$

for a parabolic profile (hood impact) using

$$\bar{a} = \frac{v_{head\perp}^2}{4s_{dyn}} \quad ; \quad t_1 = \frac{4s_{dyn}}{v_{head\perp}} \qquad (2).$$

8. *Head loadings during road impact:* For the estimation of the head acceleration caused by impact on the pavement, the head impact velocity $v_{head\perp}$ was calculated from falling height h of the pedestrian, the throwing velocity v_0 and the throwing angle α leaving the vehicle:

$$v_{head\perp} = \sqrt{v_0^2 \cdot \sin^2 \alpha + 2gh} \qquad (3).$$

Maximum dynamic cranial deformation was assumed to be s_{dyn} = 0.003 m in the case of an intact skull and 0.01 m in the case of a fractured skull, resp. [20]. For the calculation of the head acceleration Eq.(1) was used.

Case-Overview

Age, sex, height, weight, impact area on the car, and the collision speed are listed in Table 1. The head acceleration and the sustained head injuries are presented in Table 2.

Case Nb.	Age/ Sex	Height.[m]/ weight. [Kg]	Head Impact - Car -	Collision-speed v_k [Km/h]
1	79 / f	1.50 / 71.5	no contact	18-25
2	74 / m	1.68 / 58.6	air intake grill	21-23
3	80 / m	1.69 / 69.0	windshield	≈32.8
4	82 / f	1.62 / 45.7	windshield	48-61
5	47 / m	1.71 / 58.6	windshield-wiper	50-55
6	68 / f	1.55 / 57.3	windshield	78-80
7	54 / f	1.59 / 53.1	A-pillar	60-70
8	20 / m	1.62 / 62.7	motorhood	35-40
9	61 / m	1.62 / 68.4	windshield	≈78
10	93 / f	1.60 / 49.7	windshield	80-100
11	56 / m	1.72 / 83.9	windshield	70-80
12	81 / m	1.75 / 84.0	windshield	≈50
13	44 / m	1.78 / 75.6	A-pillar	55-61.6
14	83 / m	1.73 / 91.9	windshield	40-50
15	79 / m	1.71 / 77.8	windshield	50-55
16	42 / m	1.73 / 84.6	windshield	55-60
17	51 / m	1.73 / 88.4	windshield	50-60
18	83 / m	1.67 / 81.0	windshield	55-59
19	50 / m	1.77 / 51.2	A-pillar	60-70
20	51 / m	1.78 / 67.8	windshield-frame	68-71
21	66 / m	1.67 / 73.7	no contact	60-80
22	22 / m	1.83 / 93.6	no contact	≈60
23	25 / m	1.84 / 74.7	windshield	50-55
24	48 / f	1.67 / 80.7	A-pillar	40-50
25	22 / m	1.69 / 68.1	no contact	≈70

Table 1: Case-overview - general specifications

Case Nb.	Car-Impact $\bar{a}$ [g]	Car-Impact t_1 [ms]	Pavement-Impact $\bar{a}$ [g]	Pavement-Impact t_1 [ms]	Head Injuries	Origin of Injury
1	-	-	308-269	2.2-2.5	SDH/SAH/ZBL/GZ	pavement
2	-	-	351-602	2-1.6	EDH/SDH/SAH/ZBL/GZ	pavement
3	84-42	11-22	228-200	2.5-2.9	SDH/SAH/ZBL	pavement
4	85-139	12-9.4	-	-	no injury	-
5	138-238	7.7-6.5	100-97	3.8-4.1	SDH/ZBL	car
6	68.4-145	24-19	189-140	2.3-3.2	SAH	car
7	177-260	4.8-3.3	40-30	12.2-16.3	SDH/SAH/GZ	car
8	24-57.3	41-27	146-139	2.4-2.7	SAH/ZBL	car+pavem.
9	123-159	13-16	43.4-32.5	11.9-15.8	SDH/SAH/GZ	car
10	283-442	6-4.8	112-133	4.3-3.9	SAH/ZBL/GZ	car
11	142-193	12-10	≈99	≈4.5	EDH/ZBL/GZ	car
12	91-115	6-4.5	141-123	3.2 3.6	SDH/SAH/ZBL	no decision
13	206-227	4.8-3.9	197-123	2.3-3.6	SDH/ZBL/GZ	car
14	63-94.4	≈18	117-159	3.5-3.2	SDH/SAH/ZBL	car
15	≈98.3	≈14.4	178-159	2.8-3.2	SDH/SAH	car
16	59.5-47.2	26-36	≈297	≈1.9	SAH	car+pavem.
17	98-71	14.4-24	160-114	2.5-3.5	SDH/SAH	car
18	79-91	19.6-18.3	170-121	2.4-3.4	SDH/SAH	car
19	265-260	≈3.2	98-49	4.3-9.1	EDH/SDH/SAH/ZBL/GZ	car
20	244-193	≈4.4	74-111	7.4-6.1	EDH/SDH/SAH/ZBL	car
21	140-252	12-9	-	-	SDH/SAH	car
22	-	-	200-171	2-2.9	SDH/SAH/ZBL	pavement
23	141-86	12-22	247-212	1.8-2.6	SDH/SAH/ZBL/GZ	car+pavem.
24	147-275	5.8-3.1	160-79	2.5-5	SDH/SAH/ZBL/GZ	car
25	80-40	16-32	300-150	1.8-3.7	SAH/GZ	car+pavem.

Table 2: Case overview - head accelerations and sustained injuries: $\bar{a}$ = mean head acceleration; t_1 = estimated impact duration; EDH= epidural hematoma; SDH= subdural hematoma; SAH= subarachnoidal hematoma; ZBL= contusion; GZ= brain laceration

Results and Discussion

It is commonly accepted that brain injuries which are closely connected to skull deformation and fractures (epidural and local subarachnoidal hematoma and coup contusions) are related to contact forces. The incidence of remote and diffuse injuries (subdural, intracerebral, and remote subarachnoidal hematoma and contre coup contusions) seems to depend on the level of acceleration. Linear as well as rotational accelerations or a combination of both (angular) and also shockwaves are assumed to be responsible for these injuries [11], [12], [13], [22], [24], [29].

In a retrospective analysis, however, only linear accelerations can be estimated with acceptable precision. Impact forces can not easily be estimated from accelerations: According to GOT et al. [14] the effective impact mass is considerably smaller (down to only 2 Kg) than the actual mass of the head. ENOUEN [9] has shown that the effective head mass ranges from anywhere near 50% to 150% of the actual head mass. Effective head masses lower than 70% of the actual head mass were observed at head impacts on very rigid surfaces. Masses greater than 100% due to the contributions of the neck and shoulders depend mainly on the angle of impact. It is thus to be taken into account that in estimating the force, the head mass m has to be replaced by the effective mass m_{eff}. Because so far a satisfactory way of estimating the effective mass retrospectively has not been found, a calculation of the force was omitted.

The same limitation exists in the estimation of the rotational or angular acceleration. In addition, there is the uncertainty in the determination of the center of angular motion. In the kinematics of head impact, however, pure rotational accelerations do not occur but are always connected with linear accelerations. Therefore, to a certain extent, linear accelerations serve as a measure of loading even for injuries generated more or less by rotational accelerations.

In consequence of the fact that for the head impact velocity as well as for the dynamic deformation only ranges can be assessed or reconstructed, ranges result for the mean accelerations and head impact durations. Head impact velocities and dynamic deformations are combined in a way that yields maximum ranges. For the further analysis it is essential that the actual mechanical loading is within these ranges.

THE INTEREST of the further evaluation is to find thresholds for the occurrence of the head injuries examined. For this purpose, Upper and Lower Tolerance Limits are defined as follows: The Upper Tolerance Limit (UTL) is defined by the minimum of mechanical loading above which the specified injury was always observed. The Lower Tolerance Limit (LTL) is defined by the maximum of mechanical loading below which the specified injury was never observed.

For the determination of the limits of cerebral injuries, the analysed cases are sorted in a decreasing order according to the maximum and minimum values of the estimated ranges of acceleration, as schematically shown in Fig. 1. The upper/lower threshold for a certain injury can be read off the course sorted according to the maximum/minimum values: The Upper Tolerance Limit (Fig. 1a) is given by the lowest value above which the injury in question was always observed after sorting on the maximal values. In the same way the Lower Tolerance Limit is given by the highest value below which the injury has never been observed after sorting on the minimum values (Fig. 1b).

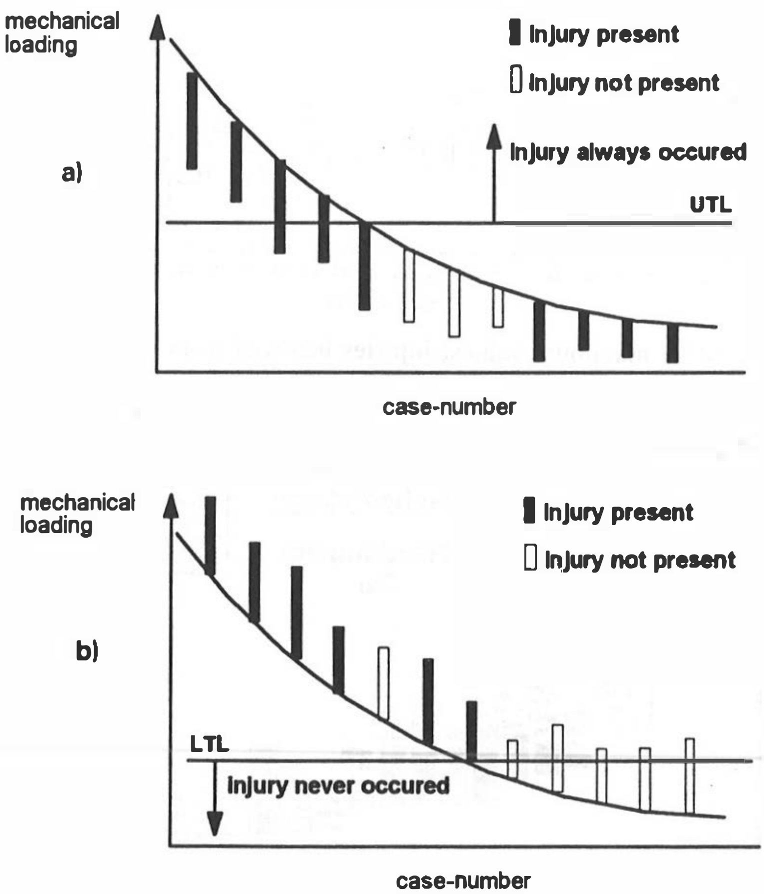

Fig. 1: Definition of Upper and Lower Tolerance Level (UTL, LTL)

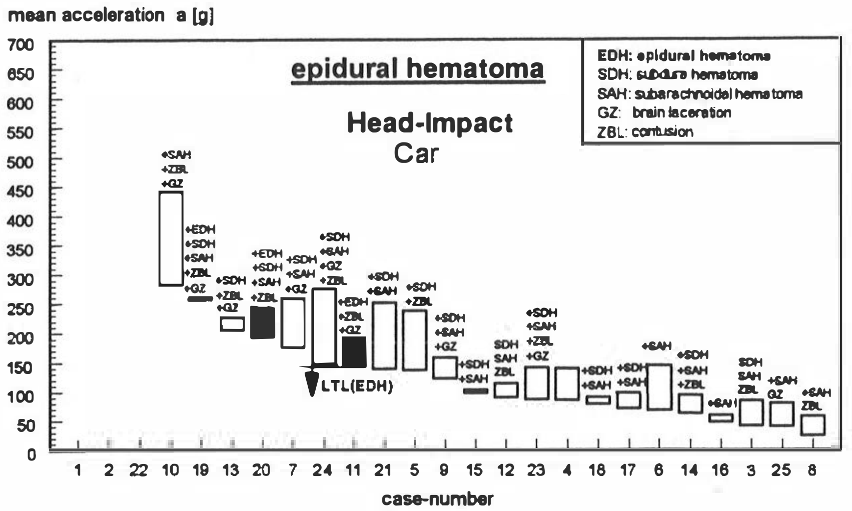

Fig. 2: Cases arranged by minimum values; injuries incurred from car are marked (+)

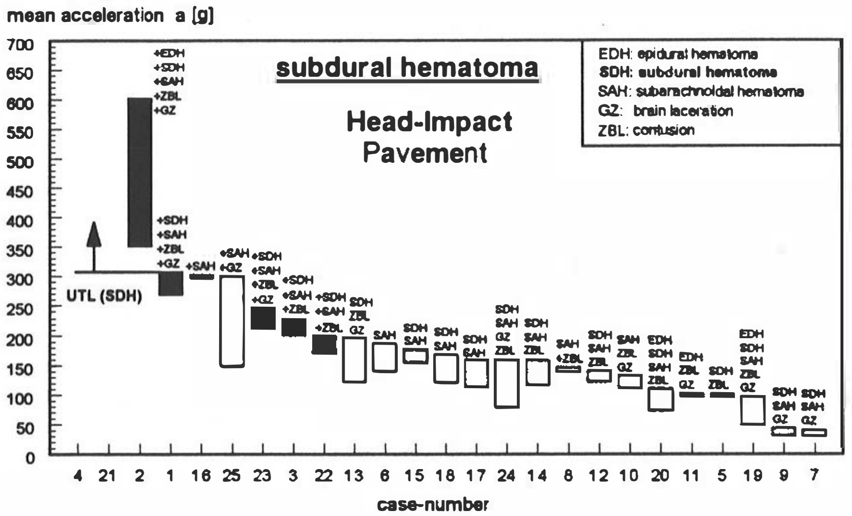

Fig. 3: Cases arranged by maximum values; injuries incurred from pavement are marked (+)

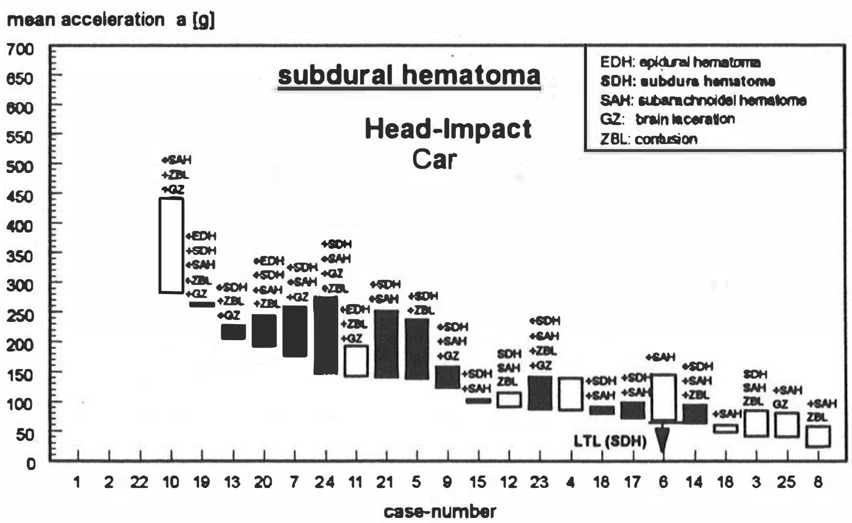

Fig. 4: Cases arranged by minimum values; injuries incurred from car are marked (+)

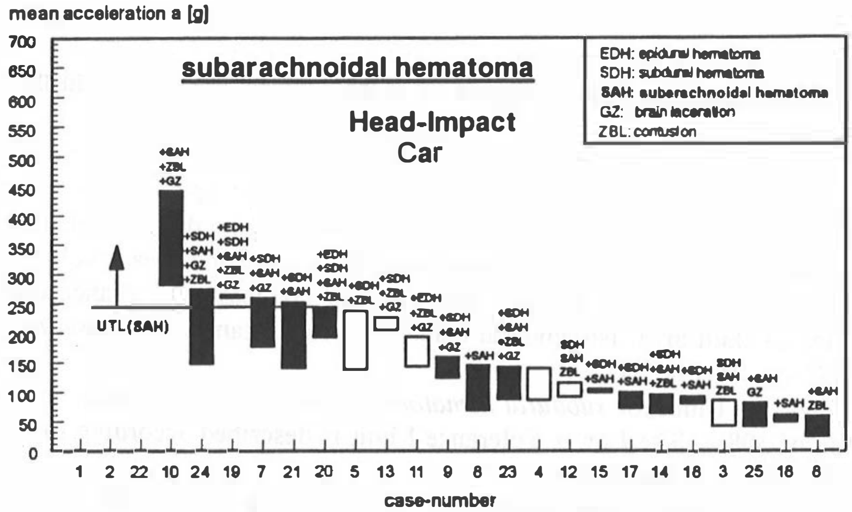

Fig. 5: Cases arranged by maximum values; injuries incurred from car are marked (+)

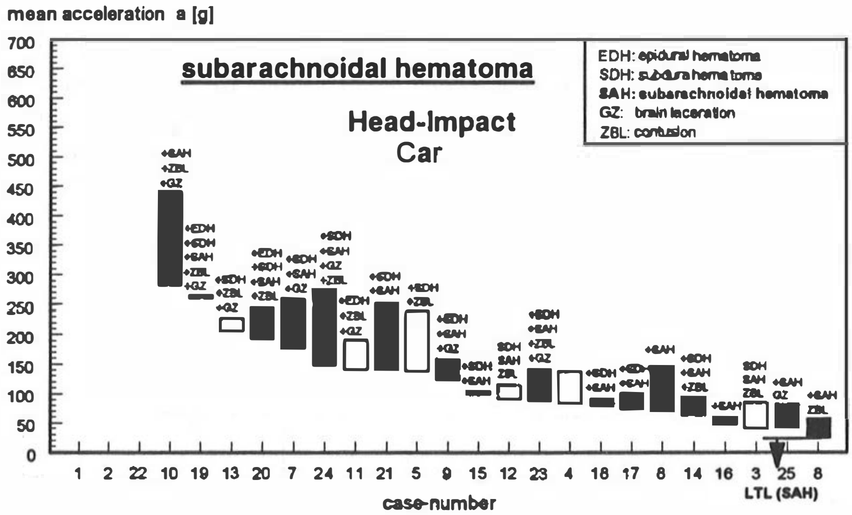

Fig. 6: Cases arranged by minimum values; injuries incurred from car are marked (+)

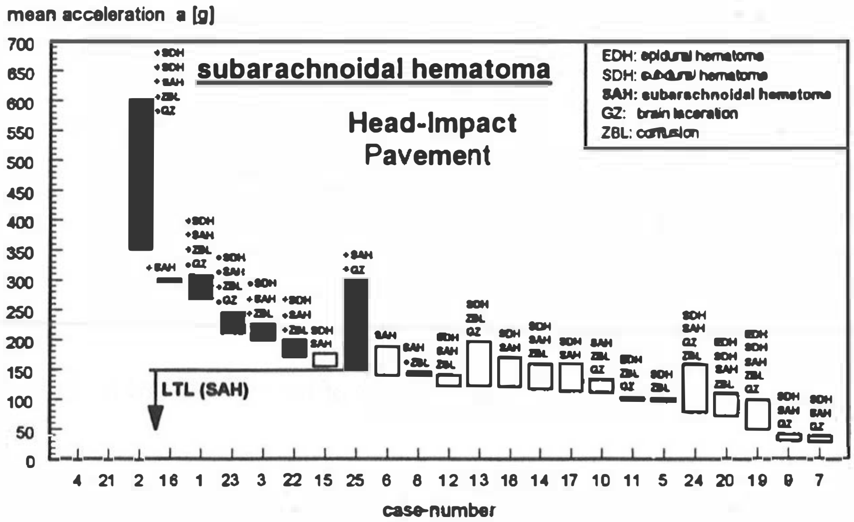

Fig. 7: Cases arranged by minimum values; injuries incurred from pavement are marked (+)

THE ESTABLISHMENT of tolerance limits for different intracranial injuries as described above is shown in Figs. 2 to 7, which are a selection out of 20 possible diagrams. Vehicle and pavement impacts are treated separately. The injuries caused by these contacts are marked in the corresponding diagrams with a (+).

The Lower Tolerance Limit for *epidural hematoma* follows from Fig. 2. The minimum mean acceleration of about 140 g (Case 11) is the lowest level at which an epidural hematoma was still observed. An Upper Tolerance Limit for epidural hematoma cannot be determined: Primarily, the impact force is responsible for this injury. Furthermore, the size of the area involved and the elastic deformation of the skull are of significance and, last but not least, the chance of a blood vessel present in the affected skull area. No epidural hematoma, for instance, was observed even in a case of at least 280 g (Case 10, Fig. 2).

The Upper Tolerance Limit for *subdural hematoma* is given by Case 1 (Fig. 3) with a maximum mean acceleration of 308 g. The Lower Tolerance Limit is described according to Fig. 4 by the car impact of Case 14 and has a value of 68 g.

In the same way a maximum mean acceleration of 250 g (Fig 5, Case 20) results for the Upper Tolerance Limit of the *subarachnoidal hematoma.* Case 25, Fig. 7 yields 150 g as the Lower Tolerance Limit in the case of short duration pulses (3-4 ms) resulting from road impact. In the case of long pulses as with the impact on the hood a value of app. 24 g at 41 ms was determined (Fig. 6, Case 8).

Cerebral hematoma and *cerebral contusions* can be observed as coup and as contre coup injuries as well. Coup injuries are caused by forces, contre coup lesions by accelerations. The threshold of coup injuries is obviously higher than that of contre coup injuries [33] which, therefore, determine the risk of injury. For contre coup contusions the Upper Tolerance Limit is found to be at 300 g, the Lower Tolerance Limit at 60 g. For *brain laceration* the corresponding values are 300 g and 90 g, resp.

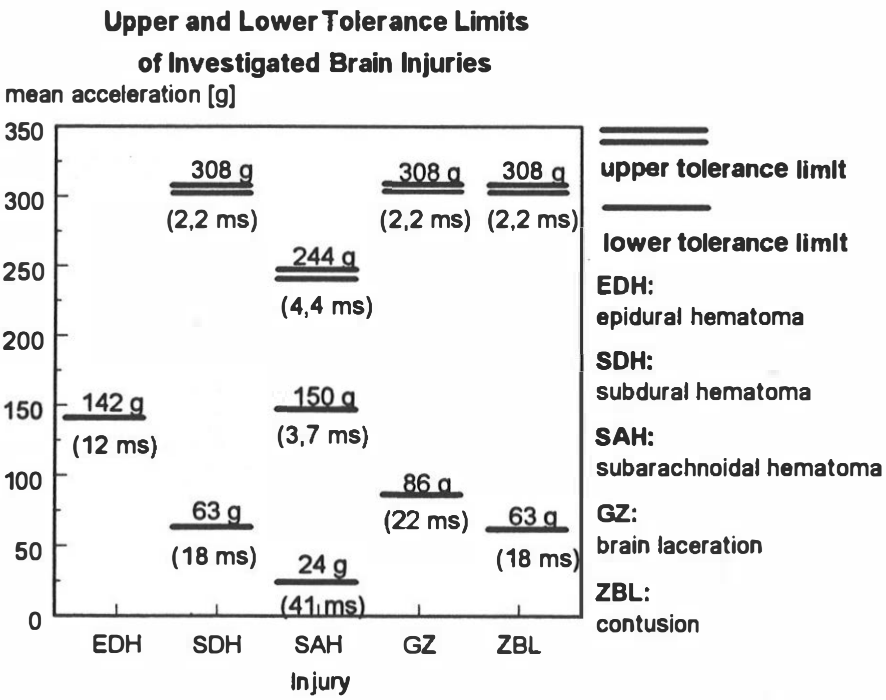

Fig. 8: Summary of upper and lower tolerance limits of the investigated brain injuries

THE FINDINGS are summarised in Fig. 8. Because mechanical tolerance limits are dependent on the pulse duration, the acceleration values in Fig. 8 are connected with impact duration. In the case of short duration impacts subarachnoidal hematoma must be expected above a mean acceleration of 250 g, above 300 g subdural hematoma, contusion, and brain laceration as well.

The values determined for the subdural hematoma are in accordance with investigations by MATTERN et al. [19] and MANAVIS et al. [18]. For the subarachnoidal hematoma similar values for cadavers and animals are reported by GENNARELLI et al. [10], NAHUM and SMITH [21], NUSHOLTZ et al. [24], STALNAKER et al. [29], and UNTERHARNSCHEIDT [32].

Because of the accumulation of errors in estimating the mean acceleration, very low values are determined for the Lower Tolerance Limits. In the case of longer lasting pulses (18-40 ms) these range between 20 and 80 g for the subdural and subarachnoidal hematoma, brain lacerations, and for contusions. We want to point out that the reverse conclusion is not allowed: These results do not imply that above these accelerations the considered injuries may occur, but merely that these are save values below which the injuries have not been observed.

CONCERNING THE age distribution of our sample (Fig. 9), 19 individuals out of 25 are above an age of 45 years. This might imply, that the results apply mainly to older individuals.

However, from our material, there is so far no indication for a significant influence of age on the tolerances of the injuries under consideration. Among those injured at rather low accelerations there are individuals of all age groups. Final conclusions have to be reserved for further investigations on larger samples.

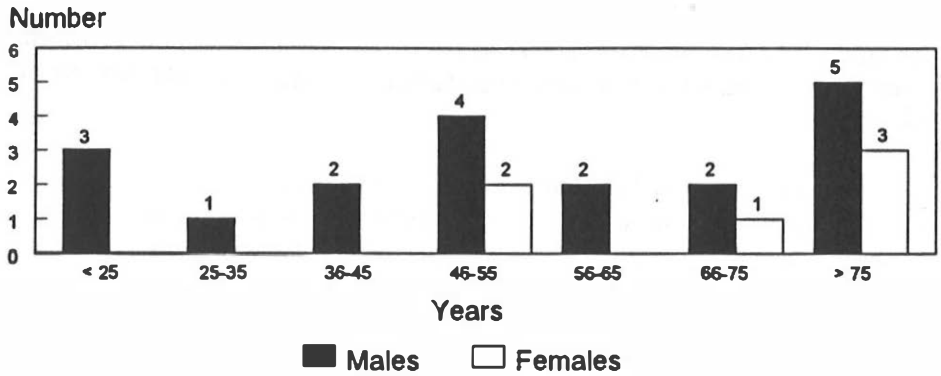

Fig. 9: Age distribution of the investigated sample

References

[1] Allsop DL, Warner CY, Wille MG, Schneider DC, Nahum AM: *Facial Impact Response - A Comparison of the Hybrid III Dummy and Human Cadaver.* Proceedings of the 32nd Stapp Car Crash Conference, Paper 881719, 139-156, Atlanta 1988

[2] Appel H, Cesari D, Tarrière C: *Joint Biomechanical Research Project KOB (Forschungsverbund Biomechanik).* Unfall- und Sicherheitsforschung Straßenver-kehr 34 (1982)

[3] Ashton SJ, van Wijk J, Cesari D: *Experimental Reconstruction and Mathematical Modelling of Real World Pedestrian Accidents.* Proceedings of the International Congress and Exposition, Paper 830189, Detroit 1983

[4] Beier G: *Anstoßgeometrie und Verletzungsbild beim tödlichen Fußgänger-Pkw-Unfall.* Beitr. gerichtl. Med. 31, 65-72 (1973)

[5] Beier G, Pfriem D: *Durch die Anstoßgeometrie bedingte Besonderheiten im Verletzungsbild tödlich verunglückter Fußgänger.* Beitr. gerichtl. Med. 32, 73-77 (1974)

[6] Beier G: *Sektionsbefunde als Grundlage verletzungsmechanischer Forschung und Begutachtung*. Hefte Unfallheilk. 181, 681-685 (1986)

[7] Burg H, Rau H: *Handbuch der Verkehrsunfallrekonstruktion*. Kippenkeim: Verlag Information Ambs GmbH 1981

[8] Danner M, Halm J: *Technische Analyse von Verkehrsunfällen*. München: Kraftfahrzeugtechnischer Verlag 1981

[9] Enouen SW: *The Development of Experimental Head Impact Procedures for Simulating Pedestrian Head Injury*. Proceedings of the 30th Stapp Car Crash Conference, Paper 861888, 199-218, San Diego 1986

[10] Gennarelli TA, Thibault LE, Ommaya AK: *Phatophysiologic Responses to Rotational and Translational Accelerations of the Head*. Proceedings of the 16th Stapp Car Crash Conference, 296-308, Detroit 1972

[11] Gennarelli TA, Abel JM, Adams H, Graham D: *Differential Tolerance of Frontal and Temporal Lobes to Contusions Induced by Angular Acceleration*. Proceedings of the 23rd Stapp Car Crash Conference, 561-586, San Diego 1979

[12] Gennarelli TA: *The State of the Art of Head Injury Biomechanics*. Proceedings of the 29th American Association For Automotive Medicine Conference, 447-463, Washington, D. C., 1985

[13] Gennarelli TA, Thibault LE, Tomei G, Wiser R, Graham D, Adams J: *Directional Dependence of Axonal Brain Injury due to Centroidal and Non-Centroidal Acceleration*. Proceedings of the 31st Stapp Car Crash Conference, 49-53, New Orleans 1987

[14] Got C, Patel A, Fayon A, Tarrière C, Walfisch G: *Results of Experimental Head Impacts on Cadavers: The Various Data Obtained and Their Relations to Some Measured Physical Parameters*. Proceedings of the 22nd Stapp Car Crash Conference, 57-99, Ann Arbor 1978

[15] Janssen E, Wismans J: *Evaluation of Vehicle-Cyclist Impacts through Dummy and Human Cadaver Tests*. Proceedings of the 11th International Technical Conference on Experimental Safety Vehicles. Washington, DC, 1987

[16] Kallieris D, Schmidt Gg: *New Aspects of Pedestrian Protection Loading and Injury Pattern in Simulated Pedestrian Accidents*. Proceedings of the 32nd Stapp Car Crash Conference, 185-196, Atlanta 1988

[17] Kühnel A: *Der Fahrzeug-Fußgänger-Unfall und seine Rekonstruktion*. Dissertation TU Berlin 1980

[18] Manavis J, Blumbergs P, Scott G: *Brain Injury in Falls causing Death*. Proceedings of the IRCOBI Conference, 77-88, Berlin 1991

[19] Mattern R, Schueler F, Schmidt Gg: *Dynamic Fronto-Occipital Head Loading of Helmet Protected Cadavers*. AGARD Conference Proceedings No. 322, Ref. No. 3, Cologne, Germany (1982)

[20] Messerer O: *Elasticität und Festigkeit der Knochen*. Stuttgart: J. G. Cotta, 1880

[21] Nahum A, Smith R: *An Experimental Model for Closed Head Impact Injury.* Proceedings of the 20th Stapp Car Crash Conference, 783-814, Dearborn 1976

[22] Newman J: *A Generalized Acceleration Model for Brain Injury Threshold (GAMBIT).* Proceedings of IRCOBI Conference, 121-131, Zürich 1986

[23] Niederer PF, Schlumpf MR: *Influence of the Vehicle Front Geometry on Impacted Pedestrian Kinematics.* Proceedings of the 28th Stapp Car Crash Conference, 135-147, Paper 841663, Chicago 1984

[24] Nusholtz GS, Lux P, Kaiker P, Janicki MA: *Head Impact Response- Skull Deformation and Angular Acceleration.* Proceedings of 28th Stapp Car Crash Conference, 41-74, Chicago 1984

[25] Pfriem D: *Verletzungsmuster und Beschädigungsmuster beim Fußgänger-Pkw-Unfall.* Dissertation Universität München 1975

[26] Ryan GA, McLean AJ, Vilenius ATS, Kloeden CN, Simpson DA, Blumbergs PC: *Head Impacts and Brain Injury in Fatally Injured Pedestrians.* Proceedings of the IRCOBI Conference, 27-37, Stockholm 1989

[27] Schneider H, Beier G: *Experiment and accident: Comparison of Dummy Test Results and Real Pedestrian Accidents.* Proceedings of the 18th Stapp Car Crash Conference, 29-69, Ann Arbor 1974

[28] Schuller E, Beier G, Spann W: *Kriterien für die Prüfung von Schutzhelmen für motorisierte Zweiradfahrer.* Abschlußbericht zum Forschungsprojekt BAST FP 2.8718, Institut für Rechtsmedizin der Universität München, München 1992

[29] Stalnaker RL, Melvin JW, Nusholtz GS, Alem NM, Benson JB: *Head Impact Response.* Proceedings of the 21st Stapp Car Crash Conference, 303-335, New Orleans 1977

[30] Stürtz G: *Experimental Simulation of the Pedestrian Impact.* Proceedings of the 10th International Technical Conference on Experimental Safety Vehicles, Paper 856120, Oxford 1985

[31] Trosseile X, Chamouard F, Tarrière C: *Reconsideration of the HIC, Taking into Account the Skull Bone Condition Factor (SBCF) - Limit of Head Tolerance in Side Impacts.* Proceedings of the 32nd Stapp Car Crash Conference, Paper 881710, 29-34, Atlanta 1988

[32] Unterharnscheidt FJ: *Translational Versus Rotational Acceleration - Animal Experiments with Measured Input.* Proceedings of the 15th Stapp Car Crash Conference, 767-771, Coronado 1971

[33] Unterharnscheidt FJ: *Die traumatischen Hirnschäden. Mechanogenese, Phatomorphologie und Klinik.* Z. Rechtsmedi. 71, 153-221 (1972)

[34] Walz F, Niederer P, Kaeser R: *Auto-Fußgängerkollision; Verletzungsreduktion - Unfallrekonstruktion - mathematische und experimentelle Simulation; Kopfverletzungen bei Zweiradkollisionen.* Interdisziplinäre Arbeitsgruppe für Unfallmechanik Universität und ETH Zürich, Zürich: 1985