Modelling and specifications for an improved helmet design

Modelling and specifications for an improved helmet design.

A literature review.

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Summary

These are the results of a literature survey, performed to investigate the current standard of helmet research.

Annually, approximately five thousand motorcyclists get killed as a result of traffic accidents. They account for 9% of all road fatalities. The majority of collisions, particularly causing head injury, are head-on impacts. Although the majority of these collisions are with cars, most serious head injuries are caused by the head hitting the road or roadside furniture. Wearing a helmet reduces the risk of fatality to about a half.

In accidents, the human head is exposed to loads exceeding several times the loading capacities of its natural protection. Mechanisms causing head injury are still not clearly understood. The response of the brain to loading of the skull may be frequency dependent and this may explain the differences in injuries found after long duration and short duration impacts.

Various injury criteria for the head have been proposed in the past. The most commonly acknowledged and widely applied head injury criterion is the HIC (Head Injury Criterion), which is based on the assumption that the linear acceleration of the head is a valid indicator of head injury thresholds. It does not take into account rotational acceleration, head kinematics nor direction of impact, even though rotational acceleration is believed to be the cause of several head injuries (e.g. Acute Subdural Haematoma).

The development of mathematical models is vital to a better understanding of the various head injury mechanisms. Lumped mass models have provided insight into the simple behaviour of the head and brain, but finite element modelling is the only method that can predict intracerebral parameters. It also allows for complex anatomical structures to be modelled. However, a correct human head model is very much dependent on the validity of the material models of the modelled structures.

The exact manner in which helmets protect the head is still not understood. Current helmets are empirically designed to meet the shock absorption requirements of current test standards. Again, lumped mass models were very useful in parametric studies, but to investigate the way in which a helmet protects the head it is necessary to use 3D finite element modelling. The energy absorbing capacities of a helmet were found to be very much dependent on the material parameters of the protective padding liner.

Current standards all require tests for impact energy absorption, but most of them do not require tests for the chin guard or resistance to penetration. Efficient energy absorption with a minimum tendency to induce rotational motion and a comprehensive evaluation of the whole helmet including the chin guard of a full-face helmet are features which require special attention.

Chapter 1

Introduction

The human head and neck system is considered to be the most critical body region in crash situations, because of the often irreversible nature of injuries to the central nervous system. Furthermore, the head is identified as the body area most frequently involved in life-threatening injury [ETSC, 1993]. Brain injuries are the most important head injuries, but probably the least well understood. However, there is considerable literature published, which has been examined to determine how and what injuries occur in motorcycle accidents.

In car crashes, the car occupants are protected by the car body itself, padding of the car interior, safety belt systems, airbags and retracting steering systems. Motorcyclists involved in car crashes are much more vulnerable than car occupants, even at relatively low speeds,with 80% of all injuries sustained at speeds of 40 km/h or less [Thomas & Bradford, 1992]. And yet almost the only protection afforded to a motorcyclist is the helmet he wears. In a study of motorcyclists injured in road accidents in Germany, Otte *et al.* [1984] report that 70% of nonhelmeted riders receive head injuries whereas this helmet provides protection for the motorcyclist's head, nevertheless a fundamental understanding of the way in which a helmet performs its protective function is still lacking. This hampers further optimisation of motorcycle helmets, suggesting the need for structured scientific research.

One way of gaining more insight in the way helmets protect the head is by means of numerical simulation. Once a functioning and validated numerical helmet model is created, one can easily perform a parametric study, and try and improve the protective function of the helmet. Such a model must at least be a 3D Finite Element Model, to account for shell vibrations and to be able to use complex material models.

Chapter 2

Accidents and injury mechanisms

2.1 Introduction

An understanding of the ways in which the helmeted head is loaded during motorcycle accidents and knowledge of the causes of head injuries is very helpful to improve helmet protection. National statistics present an overall summary of the extent of the problems arising from accidents but with little detail. In particular, injuries are generally classed as fatal, serious and slight with serious being used when the victim has spent at least one night in hospital. The records are compiled by police officers who are not medically qualified and even this broad classification leads to different interpretations. Accident studies are used, therefore, to provide the necessary information. Where injuries are classified by medical staff, the severity is usually indicated by means of the Abbreviated Injury Scale (AIS) ranging from slight at AIS 1 to almost certainly fatal at AIS 6.

Measurement of injuries using the AIS system alone does not indicate injury details nor the consequences for the casualty.

2.2 Casualty rates

The motorcycle casualty rates for many European countries together with those for other developed countries are examined. Table 2.1 shows the trends in motorcyclist fatalities, and in the proportion of all road deaths [1] accounted for by motorcyclists, in a number of countries for which statistics were available. Fatalities are declining in most countries (column 4), though there are sizable increases in Finland, Greece, Spain and Switzerland, and outside Europe in Japan and New Zealand. However, fatalities among other types of road users are declining more quickly, so that motorcycle fatalities form an increasing proportion of all road deaths in most countries (column 7). The proportion of all road fatalities is not large (9% - column 6), though substantial, illustrating the fact that motorcycling is a minority mode, but it is nevertheless sizable in absolute terms, with some 8.6 million machines owned (excluding mopeds) in the 15 European countries, and some 5 thousand fatalities annually [EEVC, 1993].

This increasing proportion of casualties should be seen against an increasing trend in motorcycle ownership in most countries, with a 21 percent increase in the European countries

[1] The table has concentrated on fatalities, because this is the most serious category of injury, and is least beset by problems of definition. In general, each death is accompanied by ten to fifteen times as many serious injuries, and thirty to fifty times as many slight injuries.

Table 2.1: Motorcyclist fatalities in various countries [EEVC, 1993].
(for motorcycles and scooters except where indicated)

Country	Number of Fatalities			As a of age of all road fatalities		
	1980	1990	% change	1980	1990	% change
Australia*	442	263	-41	13.5	10.5	-16
Austria	106	107	+1	7.3	7.2	-1
Belgium	170	106	-38	7.5	5.4	-24
Denmark	59	39	-34	8.3	6.2	-28
Finland	21	28	+33	3.1	4.3	+13
France	1136	1031	-9	8.6	9.2	+8
Germany	1232	769	-38	10.5	9.6	-8
GB	1113	621	-44	17.8	11.7	-34
Greece	106	183	+73	7.7	13.4	+73
Ireland	48	41	-15	8.5	8.6	+1
Italy	822	706	-14	9.0	10.0	+11
Japan	1163	1920	+65	9.6	13.2	+37
Netherlands	130	72	-45	6.5	5.2	-20
New Zealand	91	114	+25	15.2	15.6	+3
Norway	29	25	-14	8.0	7.5	-6
Spain	316	792	+151	4.8	8.8	+81
Sweden	43	46	+7	5.1	6.0	+18
Switzerland	139	160	+15	11.2	16.8	+50
USA	5079	3173	-38	9.8	7.0	-29
Europe #	5470	4726	-14	Av 8.3	8.7	+5
Total	12245	10196	-17	Av 9.1	9.3	+2

* Statistics include mopeds. # 15 countries above

Data from UNECE Statistics of Road Traffic Accidents in Europe, with fatalities adjusted to standard 30 day definition by the Institut für Zweiradsicherheit, and from the OECD IRTAD database.

for which data are available. The fatality rate per registered motorcycle has fallen considerably overall, by an average of 28 percent in the European countries, though there have been substantial increases in the fatality rates in Greece and Spain, and also in Japan and New Zealand. This general improvement is pleasing, but it should be noted that the OECD database also shows that car ownership has grown even more rapidly than motorcycle ownership over this period, so that the improvement in motorcycle safety has not been as great as that for road traffic generally. Overall, in 1990, the motorcycle death rate per vehicle in Europe was four times as large as that for cars. This figure is an underestimate of the higher risk of motorcycles, since it takes no account of the smaller distance travelled per year, on average, by motorcycles than by cars.

2.3 Accident causes and configurations

Table 2.2 collects together summaries of obstacles hit from four different studies, by Harms

Table 2.2: Frequency of objects struck.
(Percentage of accidents in sample*)

Study	Sample size	Collision with				
		Car	Other 4 wheel	Other m/c	Fell off	Other obstacle
Harms [1981]	766	41	6	2	30	6
Hurt Jr. *et al.* [1981]	900	65	5	3	19	6
Kalbe *et al.* [1981]	123	57	6	2	18	1
Otte *et al.* [1981]	272	68		-	18	14

* Percentage do not sum to 100 because of cases where object was not known.

[1981], Hurt Jr. *et al.* [1981], Kalbe *et al.* [1981] and Otte *et al.* [1981]. There is a general consistency, though Harms finds rather fewer cars struck, and rather more cases of the rider simply falling off, than the other studies. It is clear that the object struck most frequently, in half to two thirds of collisions, is a car. Vallée *et al.* [1981] showed that although the highest proportion of collisions are with cars, in fact they account for only 33% of the objects struck by the rider's head. This because the rider's head often does not strike the collision vehicle, but the trajectory of the rider after collision brings the head into contact with other objects, often the road, motorcycle or roadside furniture.

Chin *et al.* [1989] found that head injuries mainly occur in impacts 'head on' to the motorcycle. According to Sporner *et al.* [1989] the majority of collisions are head on. Also, Otte *et al.* [1981] and Whitaker [1980] conclude that over 80 percent of impacts are within +/-20 degrees of the front, while Harms [1989] finds 72 percent within +/- 15 degrees.

As with all vehicle collisions, the risk and seriousness of all types of injury increases with impact speed. A number of studies suggest that the mean motorcycle speed is not very high, in the range 30 to 45 km/h [eg. Hurt Jr. *et al.*, 1981; Otte *et al.*, 1981; Whitaker, 1980; ONSER, 1983; Fuller & Snider, 1987]. Whitaker [1980] and Fuller & Snider [1987] reported that speeds of the other vehicles were generally substantially less than the motorcycle speed. A substantial proportion of serious injuries, and fatalities, occur at modest speeds where there is some hope of providing protection.

When looking specifically at statistics of head injuries, Otte & Felten [1991] found that fractures of the lower jaw and skull base occurred only at speeds over 30 km/h. Severe brain injuries of severity AIS 3, however were sustained at relative speed as low as 11 km/h. Therefore there is some hope that better protection can be afforded. At the moment, progress in research is hampered by inadequate accident data, especially on accident configurations, so more effort needs to be put into this area of investigation.

2.4 Injury mechanisms

Wismans *et al.* [1994] made a load-injury model, which schematically presents the process in which an accident leads to injury (figure 2.1).

In case of an accident, a static or dynamic load will be applied on the body considered. The biomechanical response is defined by 'any change in time of the position and shape of the human body, a body region or tissue and any physiological changes related to these

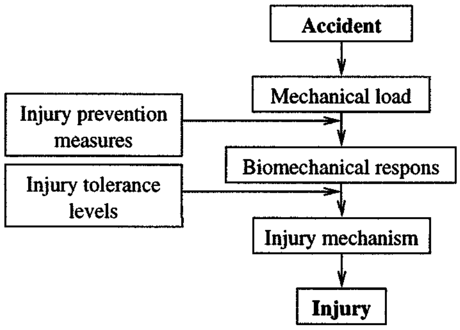


Figure 2.1: Load-injury model by Wismans *et al.* [1994].

mechanical changes'. Examples are: the brain motion and deformation after an impact but also physiological changes as dizziness, headache and changes in reflexes. Preventive measures (e.g. helmet, airbag, seatbelt, etc.) can reduce or eliminate the biomechanical response. The injury mechanism is the mechanism that causes the biomechanical response to result in injury. Injury will take place if the biomechanical response is of such nature, that tolerance levels are exceeded.

Otte *et al.* [1982] provide a detailed categorisation by body part. Their findings are fairly characteristic of findings by other accident investigators (for a summary, see [Harms, 1993]). They found that the most frequently injured parts of the body are the legs (39% of all injuries), the head (23%) and the arms (19%). However, the head injuries are more serious (at an average AIS score of 2.4) than leg injuries (AIS 1.9) or arm injuries (AIS 1.5). They account for 80 percent of the fatalities. Thus it is clear that reducing the severity of head injuries is a high priority. Hurt Jr. *et al.* [1981] found that only 39.4% of the riders and 15.9% of the passengers were wearing some kind of helmet, and it was found that the helmeted riders sustained a significantly lower injury frequency at all levels of severity.

The presence of a pillion passenger is likely to have an important effect on the outcome of a collision. In general the rider is likely to receive more severe injuries due to the load caused by the passenger's momentum, while the pillion is likely to receive less severe injuries, especially to the head, because of the cushioning effect of the rider in front, and possibly in some circumstances because the passenger is launched upwards by the back of the rider and thus is thrown over the impacted vehicle [Grandel, 1987; Otte, 1989]. However, Otte [1989] also notes, that, on the whole, injury levels to riders accompanied by pillion passengers are actually lower than those to solo riders, and he attributes this to lower average impact speeds for rider/passenger combinations than for solo riders.

2.5 Head injury patterns

This section enlarges on the area of head injuries, detailing the types of injury that are sustained by the skull, the neck and the brain. It also deals with the 'Head Injury Criterion', which is an indicator used to measure the potential for head injury.

Head injuries occur in a variety of accident configurations [Hurt Jr. & Thom, 1992], which can involve direct impact e.g. with the head striking a hard surface or a blunt object striking the head, or they can occur without direct impact, as in severe whiplash from blunt force trauma to the chest. Gennarelli [1985] has provided a biomechanical description and classification of this system of injuries, as shown in table 2.3.

Table 2.3: Mechanistic types of head injury [Gennarelli, 1985].

- □ Contact injuries (requiring impact of the head; but head motion is not necessary)
  - ◇ Skull deformation injuries
    - Local
      - *a.* Skull fractures (suture separation, indentation, linear, depressed, comminuted, crushing, massive comminution)
      - *b.* Epidural haemorrhage/haematoma EDH
      - *c.* Coup contusions, lacerations, maceration, avulsion, extrusion
    - Remote
      - *a.* Vault and basilar fractures
  - ◇ Stress wave injuries
      - *a.* Contrecoup contusion
      - *b.* Intra cerebral haemorrhage/haematoma ICH
- □ Inertial injuries (direct impact to the cranial vault is not necessary; head acceleration necessary)
  - ◇ Surface strains
      - *a.* Subdural haematoma SDH
      - *b.* Contrecoup contusion
      - *c.* Intermediate coup contusion
  - ◇ Deep strains
      - *a.* Concussion syndrome
      - *b.* Diffuse axonal injury DAI
      - *c.* Intra cerebral haemorrhage/haematoma ICH

Note that Gennarelli [1981] stated that contrecoup contusion and intra cerebral haematoma (see chapter 3) are 'most likely to be caused by concentrated strains caused by stress wave propagation', so it is not proven that stress waves are the real cause of these injuries.

Because a motorcycle helmet (to an approved standard) will spread or diffuse any contact impact force and provide for an energy absorption beneath that contact point, the contact injuries defined by Gennarelli (table 2.3) are those injuries which are most likely to be prevented — or even excluded — by a motorcycle helmet. The effectiveness of helmets in protecting the head is discussed in chapter 5, but one example of when injury can be prevented by a helmet is with the external ear. Hurt Jr. & Thom [1992] pointed out the dangers to the external ear when the rider is unhelmeted. The avulsion of the pinna is a typical injury when the unprotected head encounters pavement, sharp metal objects or other obstacles, so that motorcycle helmet coverage easily prevents such injuries.

In a study of head injuries Otte *et al.* [1984] found that injuries were predominantly located at the front side of the head. Although there were a large number of different kinetic patterns in crash and post-crash phases, the head with the face region, especially the chin and forehead, is nearly always exposed to impact risks. Approximately one third of all injuries to the helmet-protected heads of motorcyclists are minor soft part injuries, like contusions and abrasions (32.9% and 32.8% respectively). More serious soft part injuries, such as laceration/contusions, cut or scalping injuries represent another 21.9% of helmet-protected, and 25.5% of unprotected heads. Otte [1991] also found that persons who suffered a chin impact remained uninjured in only 37%, while persons with impact of the helmet and without chin impact sustained no injury in 70.1% of the cases. Otte's injury analysis shows that as a rule persons with chin impact suffer soft-part injuries three times as often (49.3% of the persons), twice as many fractures (18.1% of the persons) and twice as many skull-brain injuries (39.9% of the persons) as those without impact. Fractures to the base of the skull, lower jaw, upper jaw and top of the skull were especially frequent with chin impacts.

Otte [1991] showed that fractures sustained by the skull with chin impacts are more than twice as frequent as those sustained otherwise. This is true for fractures of all parts of the skull and facial bones, except for cheek bones. He also found that an oblique frontal impact (case II - figure 2.2) and a more sagittal impact from the front, rectangular to the face (case I) produce different injury patterns, with fundamentally different impact kinematics, although soft-part injuries as well as fractures are characteristic for both types of chin impacts. The difference in kinematics can be explained from an anatomical point of view. In case of the oblique frontal impact, more force is transmitted by the denture.

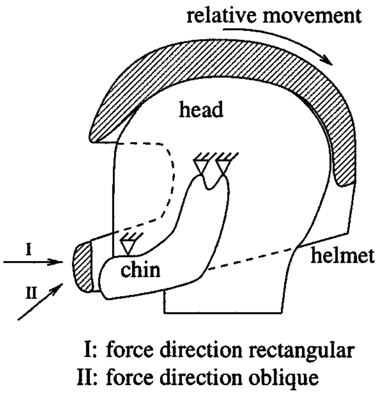


Figure 2.2: Mechanical model of the chin impact with possible force transmission to the skull [Otte, 1991].

In a study by Hurt Jr. *et al.* [1986] it was found that left side impact produced the extreme of right lateral flexion and left side basilar skull fractures and vice versa. The chin guard is an area of the helmet that requires particular attention, because a high proportion of the fatalities with head injuries sustained a fracture of the base of the skull, caused by a direct impact through the chin guard to the facial skull.

2.6 In conclusion

In total there are some 8.6 million motorcycles (not including mopeds) in the 15 European countries, and approximately five thousand fatalities annually, accounting for a substantial proportion (9%) of total road fatalities. Thus, although motorcycling is a minority mode in most countries, the more exposed position of the rider in comparison with the occupants of a four-wheeled vehicle carries a risk of injury that is considerably higher than for car users; by a factor of ten per kilometre travelled, for the European countries.

The majority of collisions, particularly those causing head injury, are head-on impacts. Although the majority of these collisions are with cars, most serious head injuries are caused by the head hitting the road or roadside furniture. Most motorcycle collisions take place at relatively low speeds of around 30 km/h. Skull fractures occur at speeds of 30 km/h upwards, but brain injuries can be sustained at relative speeds as low as 11 km/h.

Head injuries occur in a variety of different accident configurations, involving direct or indirect contact. A considerable number of these injuries could be prevented or reduced by the use of a helmet and, more importantly, by improvements to existing helmets. Most head injuries are sustained at the front of the head, with more than two thirds of skull fractures involving chin impact.

A high proportion of fatalities with head injuries sustained a fractured base of the skull which was almost always caused by a direct impact, through the chin guard, to the facial skull and in turn through to the skull base. Thus the chin guard is an area of the helmet that requires particular attention.

Chapter 3

Biomechanics of head injury

3.1 Introduction

In accidents, the human head is exposed to loads exceeding, several times, the loading capacities of its natural protection. This explains why, despite extensive research, head injury is still by far the most devastating disease afflicting humanity.

Mechanisms causing head injury are still not clearly understood. Traditionally, head injuries have been related to impacts and accelerations of the head and research has concentrated upon the effects of these two types of loading. Usually, impact and acceleration were studied for their ability to cause only few particular head injuries. However, in traffic accidents impact and acceleration are inseparable and a wide range of head injuries occur.

3.2 Head injuries

Head injuries can be divided into cranial injuries (skull fractures) and intracranial injuries (injuries to vascular and neurological tissue). The term head injury comprises various kinds of trauma to the skull and its contents. Usually, several different types of head injury occur simultaneously in a traffic accident. The anatomical location of the lesions and their severity determine the physiological consequences. Figure 3.1 shows the main anatomical structures of the head and their locations inside the head.

3.2.1 Cranial injuries

Skull fractures can occur with and without brain damage, but is in itself not an important cause of neurological injury [Gennarelli, 1985; Prasad *et al.*, 1985]. Skull fracture can be either open or closed. A closed fracture is a break in the bone, but with no break of the overlying skin. An open fracture, on the other hand, is a contiguous break in both the skin and underlying bone and is more serious than a closed fracture, because of the accompanying risk for infections.

Fractures to the neuro-cranium are divided into basilar skull fractures and vault fractures (fractures to the non-base part of the skull). Basilar fractures are considered clinically significant, because the dura may be torn adjacent to the fracture site and thus highly increase the probability of contamination of the central nervous system.

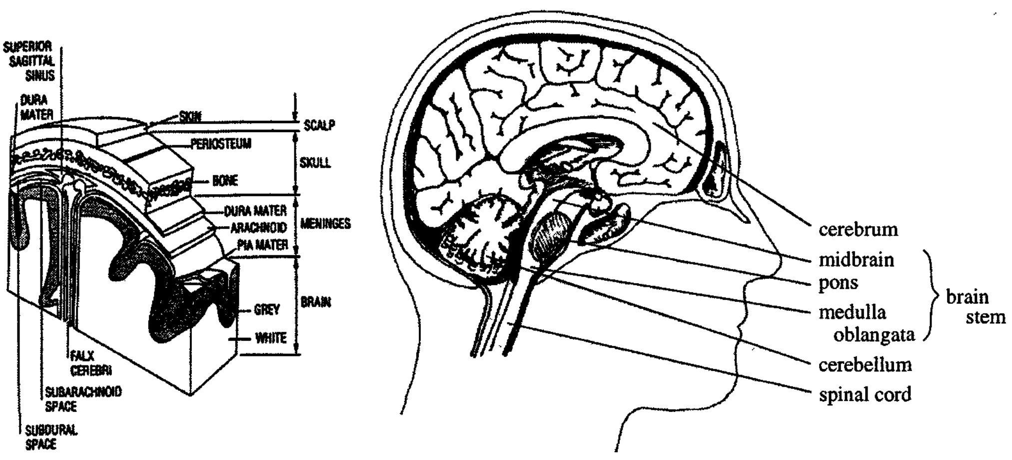


Figure 3.1: Main anatomical structures of the head and their locations inside the head.

Vault fractures are divided into linear and depressed fractures. Linear fracture (no bone displacement) is considered not severe (max. AIS 2) and does not have much significance on the course of brain injury, although this subject is still controversial [Wismans *et al.*, 1994]. Depressed fractures (with bone displacement) are likely to be associated with neural injury and/or intracranial haematoma, especially when the depression is deeper than the thickness of the skull [Prasad *et al.*, 1985].

3.2.2 Intracranial injuries

Various types of brain injury may occur due to impact. Generally two categories are distinguished: diffuse injury and focal injury. Diffuse brain injury accounts for ca. 40% of head injury patients admitted to hospitals and comprise ca. 1/3 of the deaths. Focal brain injuries account for approximately 50% of all patients admitted to hospitals and are responsible for 2/3 of head injury deaths.

Diffuse brain injury

Diffuse brain injuries range from mild concussion (without anatomical disruption of the brain) to diffuse white matter shearing injury (with anatomical disruption). In order of increasing severity, the most important types of diffuse brain injury are discussed below [Gennarelli, 1981].

- Mild concussion includes those types of brain injury resulting in confusion, disorientation and/or minor loss of memory. This type of injury does not involve loss of consciousness, and is completely reversible.

- Classical cerebral concussion involves temporary loss of consciousness which lasts less than 24 hours and is reversible.

- Diffuse white matter shearing injury (DWSI) or diffuse axonal injury (DAI) is an extreme form of diffuse brain injury with prolonged loss of consciousness (more than 24 hours) and brainstem disfunction.

Focal brain injury

Focal brain injuries are those in which a lesion has occurred large enough to be visualised without special equipment (provided an autopsy would be possible), thus always include anatomical damage. Four types of focal brain injuries are distinguished:

- Epidural haematoma (EDH), directly resulting from skull deformation, are usually associated with skull fracture and concern the meningeal vessels directly underneath the skull. EDH has a low incidence and is therefore considered of minor clinical relevance.
- Subdural haematoma (SDH) of which the acute form (ASDH) is the most severe. This type of brain injury is of high clinical relevance, especially because of the poor outcome: most studies report a mortality rate which exceeds 35%. The most important cause of ASDH is tearing of the bridging veins and arteries, crossing the subdural space.
- Contusion, the most frequently found trauma following head impact, occurs at the site of impact (coup contusion) or at remote sites of the impact (contre-coup contusion). Mortality rates reported for this type of injury range from 25% to 60% with a tendency to increase with increasing age.
- Intracerebral haematoma (ICH) include homogeneous collections of blood within the brain and are distinguished from contusions by a more pronounced localisation of the haematoma. Mortality rates reported differ a lot (6% to 72%) and survivable outcome is considerably affected by the presence or absence of loss of consciousness.

3.3 Head injury mechanisms

3.3.1 Dynamics of impact

Contact impact causes a great variety of mechanical effects to the head either because of contact phenomena and/or inertial effects. Contact phenomena predominantly cause focal head injuries. Another possibly important response of the head due to contact impact is the propagation of stress waves in the skull or the brain, which may cause focal injuries, distant from the site of impact (contre-coup). However, this assumption is not yet validated.

Generally, an impact to the head results in acceleration of the head, which leads to inertial loading of the intracranial structures. Accelerations can be translational (linear) and rotational (angular) which can result in concussion and diffuse brain injury rather than focal injury, with one exception of (A)SDH. Especially rotational acceleration is the most important cause for severe head injury: SDH and shearing injury [Gennarelli, 1981; Ommaya, 1988].

Figure 3.2 shows the occurrence of the most severe head injury in relationship (qualitatively) with angular acceleration amplitude and time duration of this acceleration [Wismans *et al.*, 1994]. The trend is that at short pulse durations, cerebral concussion can be produced (along with cortical contusion). But as acceleration magnitude increases, strain rate sensitive bridging veins may be torn and subdural haematoma results. At longer pulse duration, cerebral concussion can be achieved at lower acceleration levels, but it takes considerably ore acceleration to cause subdural bridging vein rupture. The incidence of cerebral contusion also further decreases with increasing pulse duration (not shown in figure 3.2). Shearing brain injuries are thought to be caused by high angular acceleration at longer pulse durations.

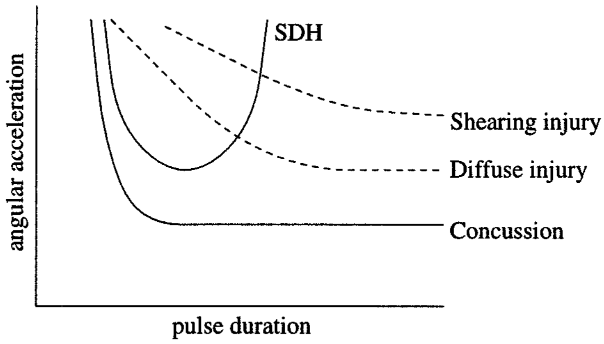


Figure 3.2: Relationship between angular acceleration and head injury [Wismans *et al.*, 1994].

3.3.2 Cranial injuries

According to Gurdjian *et al.* [1950] and Thomas *et al.* [1973], skull bending is the cause of linear skull fracture. As a result of an impact, the skull bends inwards at the site of the impact and bends outwards at some distance from the impact site. When the skull is deformed beyond its loading capacity, it fractures. Since bone is weaker in tension than in compression, cracks will appear at the skull's outer table in the regions where the skull bends outwards and on the inner table in regions in which the skull bends inwards (figure 3.3, the arrows denote the sites under tension resulting from skull bending).

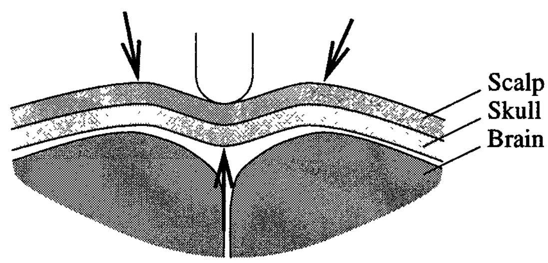


Figure 3.3: Skull bending caused by impact.

In an extensive survey of the literature on basilar skull fracture by Huelke *et al.* [1988], several mechanisms were proposed as the cause of these fractures. Originally, it was thought that basilar fracture results from cranial vault impacts, causing deformations remote from the impact site [Gurdjian *et al.*, 1949, 1953; Walker, 1973]. Thom & Hurt [1993] found that axial loading of the neck was significantly associated with basilar fracture for unhelmeted motorcyclists.

There are indications that basilar skull fractures can also be caused by impacts to the face and especially the mandible [Hodgson *et al.*, 1970; Thomas *et al.*, 1973; Brit & Herrick *et al.*, 1980; Harvey & Jones, 1980; Lau *et al.*, 1987]. Hurt Jr. *et al.* [1981] studied 900 motorcycle accidents in Los Angeles and noted that in severe impacts to the mandible, the transmission of the force through the mandible could produce a basilar skull fracture with laceration of the base of the brain.

A remarkable finding in research on basilar skull fractures is that of Alem *et al.* [1984]. They impacted the crowns of heads of unembalmed cadavers and results showed a rigid impacting surface with sufficient impact energy to cause fractures at the impact site. However, if under the same conditions the impact site was padded, the fractures appeared at the base of the skull. Increasing the thickness of the padding prevented skull fractures, but the fractures then occurred in the cervical spine.

3.3.3 Intracranial injuries

Injury caused by skull deformation

Fragments of bone resulting from skull fracture or skull penetration has been shown to cause damage to underlying meningeal and cortical tissues. The dura is adherent to the inner aspect of the cranial bones, particularly at the sutures and at the base of the skull, and contains several blood vessels. Skull deformation or skull fracture can easily cause rupture of these blood vessels, leading to an extradural haematoma [Adams *et al.*, 1980; Cooper, 1982; Chapon *et al.*, 1985]. Acute subdural haematomas can be caused by direct laceration of the bridging veins or the cortical veins and arteries by penetration wounds resulting from impacts to the head [Gennarelli, 1985]. Large cortical contusions resulting from skull deformation or skull penetration can lead to subdural haematomas [Gennarelli, 1985].

Several researchers have addressed skull denting as a cause of cortical contusions [Holbourn, 1943, 1945; Gurdjian & Gurdjian, 1976; Nusholtz *et al.*, 1984; Gennarelli, 1985]. Unfortunately, publications of experiments to validate the several points of view have not been found.

Injury caused by relative movement between the skull and the brain

The skull is smooth at the vertex, but highly irregular at the base. Therefore, sliding of the brain against the internal surface of the skull is facilitated at the vertex, but is impeded at the skull base. This can lead to high shear strains in the meningeal and cortical tissues at the skull base. Most cerebral contusions occur at the frontal and temporal lobes [Courville, 1942; Gurdjian, 1966], regardless of whether the site of impact is frontal or occipital [Gurdjian *et al.*, 1955]. On the other hand, the relative movements between the skull and the brain at the vertex lead to high strains in the structures tethering the brain to the vault of the skull. Rupture of bridging veins due to these high strains are considered to be the main cause of subdural, subarachnoidal and cortical haematomas in this area [Holbourn, 1943; Löwenhielm, 1974; Abel *et al.*, 1978; Gennarelli, 1985; Adams *et al.*, 1986].

Rotation of the skull relative to the brain presses the highly irregular skull base towards the brain (figure 3.4). This leads to a combined compression and shearing of the meningeal and cortical tissues in this area, which increases the effects of the sliding of the brain over the skull base [COST 327, 1997, chap. 3]. The effects of this relative rotation are most severe when the head is subjected to a backward non-centroidal rotational acceleration, or in case of a forward non-centroidal rotational deceleration.

The relative movement between the skull and the brain is always towards the site of impact. Because of this, intracranial tissue is compressed at the site of impact and strained at the contra lateral site. This leads to positive pressure at the site of impact and negative pressures at the opposite site (figure 3.5). This effect was clearly visible in experiments by Nahum *et al.* [1977], where pressurised cadaver heads were subjected to frontal impact.

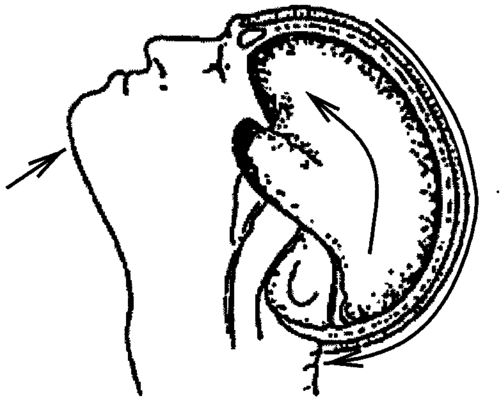


Figure 3.4: Rotation of the skull towards the brain [Sellier & Unterharnscheidt, 1963].

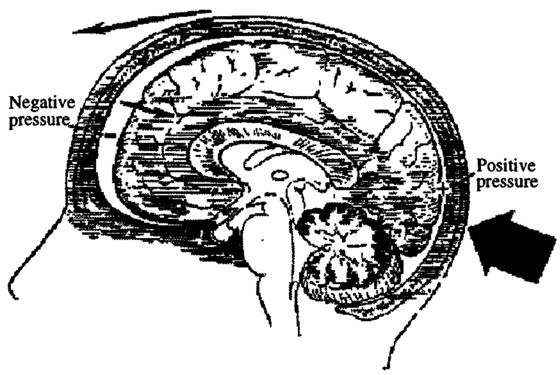
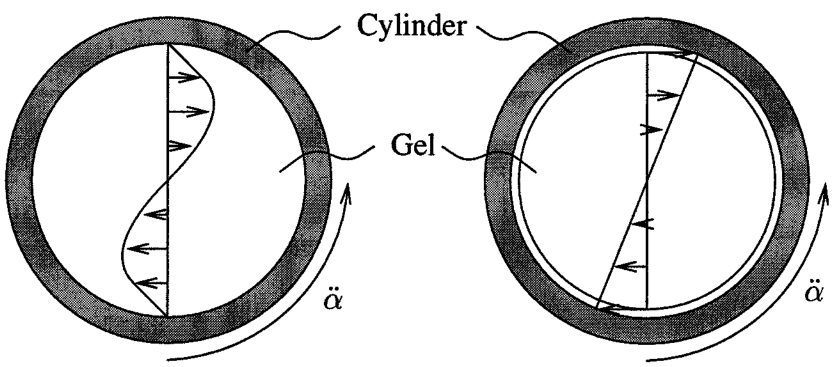


Figure 3.5: Intracranial pressure changes due to relative movement between brain and skull [Douglass *et al.*, 1968].

Injury from relative movement within the brain

The brain is inhomogeneous, it consists of various parts with different material properties. Acceleration of the head differentially loads these different parts of the brain and hence relative movement and, therefore, deformation occurs between the various parts. The brain contains several membranes (e.g., the falx and tentorium), that are much stiffer than the surrounding neurological tissues and these hinder the relative movement. This leads to considerable deformations in the brain at the contact interfaces between the brain and the membranes and is thought to be the main cause of contusions [e.g. Gennarelli, 1985].

It should be noted that in models of the human head, the modelling of the skull-brain interface is critical. If the brain is fixed rigidly to the skull, the maximum deformation in a rotational acceleration test occurs remote from the skull-brain interface (figure 3.6a). If the brain is allowed to slip relative to the skull, the maximum deformation moves more towards the skull (figure 3.6b). The human head lies between these extremes [COST 327, 1997].



a. non-slip condition b. free-slip condition

Figure 3.6: Two ideal situations representing the skull-brain interface.

3.4 Head injury criteria

For over 30 years, research has been undertaken to assess the mechanisms causing head injury in impact conditions and to establish associated tolerance levels of the human head. The development of injury criteria has been a mayor goal, in order to be able to evaluate injury prevention measures.

3.4.1 Wayne State Tolerance Curve

The Wayne State Tolerance Curve is considered to be the foundation of research on human head injury criteria. This curve evolved from the work of Lissner *et al.* [1960]; Gurdjian *et al.* [1953, 1961]; Patrick [1963], and gives tolerable average acceleration in A-P direction (Anterior-Posterior) as a function of the pulse duration. It still is the basis for the most currently accepted injury criteria. The curve is given in figure 3.7. Slight cerebral concussion without any permanent effects was considered to be within human tolerance. Only translational accelerations were used in the development of the curve which was obtained from different experiments with cadavers (I), with linear skull fracture as injury criterion (known to be highly associated with brain concussion), experiments with animals (II), where intracranial pressure was measured and compared, and experiments with volunteers (III), with loss of consciousness as injury criterion. Except for long duration accelerations, the WST-curve has never been validated for living human beings.

3.4.2 Severity Index

Gadd [1966] argued that neither the average acceleration nor the peak acceleration observed in an impact are sufficient to determine, accurately, the response of the head to an impact. According to Gadd, the resulting injury potential is highly dependent upon the acceleration pulse and therefore pulses with the same average acceleration but different shapes can have very different effects. To account for both the acceleration pulse shape and its duration, Gadd suggested integrating the acceleration signal over its entire duration. Gadd further maintained that injury potential was a non-linear function of acceleration magnitude. Therefore, Gadd suggested that an exponential weighting factor (greater than 1) be applied to the acceleration

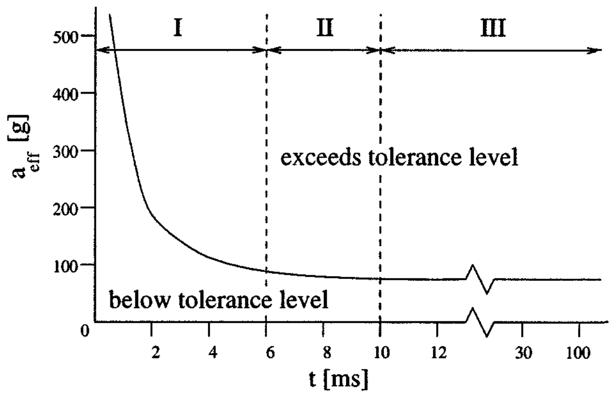


Figure 3.7: Wayne State Tolerance Curve.

and that the result be integrated over the duration of the acceleration. This led to the following injury criterion, called the Severity Index:

$$(G)SI = \int_0^T a(t)^{2.5}dt \quad (a(t) \text{ in g's}) \tag{3.1}$$

The weighting factor 2.5 only applies to the head and is primarily based on a straight-line approximation of the WSTC plotted on log-log paper between 2.5 and 50 ms. Gadd proposed a threshold (tolerance level) for concussion for frontal impact of 1000, which agreed with the WST-curve, the Eiband tolerance curve [Eiband, 1959] and the accident simulation data by Swearingen [1965]. Gadd [1971], later suggested a threshold of 1500 for non-contact loads on the head. The (G)SI has received significant scientific criticism, because it deviates considerably from the WST-curve [e.g. Slattenschek & Tauffkirchen, 1970].

The WST-curve was based on average acceleration, therefore, an approximation of this curve should also represent the average acceleration of the considered pulse. Versace [1971] suggested, therefore, the following injury criterion:

$$(V)SI = \left[\frac{\int_0^T a(t)dt}{T}\right] T \tag{3.2}$$

3.4.3 Head Injury Criterion, HIC

Based on the Versace's criticism on the (G)SI, NHTSA [1972] suggested that the SI should be replaced with a slightly modified injury criterion, called the Head Injury Criterion (HIC):

$$HIC = \left\{\left[\frac{\int_{t_1}^{t_2} a(t)dt}{t_2 - t_1}\right](t_2 - t_1)\right\}_{\max} \tag{3.3}$$

with t_1 and $t_2[s]$ any two points in time during any interval in the impact and $a(t)$ the resultant head acceleration in g's (measured at the head's centre of gravity).

As for the SI, a value of 1000 is specified for the HIC as concussion tolerance level for concussion in frontal (contact) impact. For practical reasons, the maximum time interval $(t_2 - t_1)$ which is considered to give appropriate HIC values was set to 36 ms [SAE, 1986]. This time interval greatly affects HIC calculation and recently, this time interval has been proposed to be further reduced to 15 ms in order to restrict the use of the HIC to hard contact impacts [Hodgson & Thomas, 1972].

Most important drawbacks of the HIC are that the WSTC lacks a functional relationship between human head injury and human surrogate head acceleration-time response and that HIC only takes into account the linear aspects of head motion, thus no angular accelerations are taken into account. Despite its drawbacks, HIC is the most commonly used criterion for head injury in automotive research.

3.4.4 Generalised Acceleration Model for Brain Injury Threshold, GAMBIT

The previously discussed injury criteria concern linear head impact response. In the previous section, the importance of rotational acceleration of the head was noted, especially with respect to ASDH and diffuse brain injury. A summary of various tolerances of the human brain to angular acceleration (and angular velocity) is given in table 3.1.

Table 3.1: Human brain tolerance to angular acceleration (and angular velocity) concerning sagittal head motion [Prasad *et al.*, 1985].

injury	**tolerance**
cerebral concussion	50% probability: for t $<$ 20 ms: $\ddot{\alpha}$=1800 rad/s^2 for t $\geq$ 20 ms: $\dot{\alpha}$=30 rad/s
bridging vein rupture	$\ddot{\alpha}$=4500 rad/s^2 and/or $\dot{\alpha}$=70 rad/s
brain surface shearing	2000 $< \ddot{\alpha} <$ 3000 rad/s^2
brain (general)	$\dot{\alpha} <$ 30 rad/s: safe: $\ddot{\alpha}$ <4500 rad/s^2 AIS 5: $\ddot{\alpha}$ >4500 rad/s^2 $\dot{\alpha} >$ 30 rad/s: AIS 2: $\ddot{\alpha}$=1700 rad/s^2 AIS 3: $\ddot{\alpha}$=3000 rad/s^2 AIS 4: $\ddot{\alpha}$=3900 rad/s^2 AIS 5: $\ddot{\alpha}$=4500 rad/s^2

An attempt to combine translational and rotational head acceleration response was also made by Newman [1986]. Considering these accelerations as the cause for stresses generated in the brain and resulting in brain injury, a Generalized Acceleration Model for Brain Injury Threshold (GAMBIT) is proposed. The general GAMBIT equation is:

$$G(t) = \left[\left(\frac{a(t)}{a_c}\right)^n + \left(\frac{\ddot{\alpha}(t)}{\ddot{\alpha}_c}\right)^m\right]^{1/s} \tag{3.4}$$

with

$a(t)$ and $\ddot{\alpha}(t)$ the instantaneous values of translational and rotational acceleration respectively;

n, m and s empirical constants selected to fit available data;

a_c and $\ddot{\alpha}_c$ the critical values of the accelerations (tolerances).

On the assumption that the tolerances derived from experiments with only translational or only rotational head motion are also valid for combined head response, and on the assumption that translational and rotational acceleration equally contribute to head injury, Newman simplified this equation to become:

$$G = \frac{a_m}{250} + \frac{\ddot{\alpha}_m}{10000} \leq 1 \tag{3.5}$$

with

a_m [g] and $\ddot{\alpha}_m$ [rad/s^2] the mean values of linear and angular acceleration respectively;
250 g being the maximum allowable linear acceleration (g=9.81 m/s^2);
10,000 rad/s^2 being the maximum allowable angular acceleration;

The GAMBIT, however, thus far lacks extensive validation.

3.5 In conclusion

Defining the causes of head injury is not an easy task, because several different types of head injury can originate from the same accident. Certain head injuries will have more severe consequences than others, e.g. extensive axonal damage and subdural haematoma and will, therefore, determine the overall outcome.

Injuries from an impact can occur at, or remote from, the site of contact. The effects of the impact at the site of contact are fairly well understood and are known to cause deformation, fracture and penetration of the skull (mainly the vault), whereas the effects remote from an impact are still not clearly understood.

The response of the brain to loading of the skull may be frequency dependent and this may explain the differences in injuries found after long duration (low frequency) and short duration (high frequency) impacts. However, the response of the brain to a load on the skull remains largely unknown.

Rotational and translational acceleration almost always occur together in an accident and both cause injury. As the effect of rotational acceleration is concerned, duration is thought to be critical to the outcome and research should be directed to finding the threshold of injury from rotational motion.

Various injury criteria for the head have been proposed in the past. The most commonly acknowledged and widely applied head injury criterion is the HIC, which is based on the assumption that the linear acceleration of the head is a valid indicator of head injury thresholds. This criterion has enabled vehicle safety to be improved. Nevertheless, it has shortcomings and does not take into account rotational acceleration, head kinematics nor direction of impact. Future research should be directed to the derivation of a criterion which overcomes these criticisms.

Chapter 4

Mathematical modelling of the human head

4.1 Introduction

The development of mathematical models is vital to a better understanding of the various head injuries and head injury mechanisms. This chapter reviews the development of mathematical models of the head from the basic analytical type to advanced three dimensional finite element models. Finite element models have become the most widely used mathematical models. The main limitation of finite element head models is lack of data for the characteristics of the materials of the human head particularly the brain.

4.2 Analytical head models

Analytical models simulate the skull typically by a rigid spherical shell and the brain by a fluid [Anzelius, 1943; Gross, 1958], which can be elastic [Engin, 1969] or viscoelastic [Lee & Advani, 1970; Bycroft, 1973; Margulies & Thibault, 1989], see figure 4.1. Analytical models give stress and strain distributions.

These models were very helpful for investigating possible brain injury mechanisms generated by shear stress, or wave propagation under linear and angular acceleration. This kind of model was limited mainly by the geometry, which was very much idealised.

4.3 Lumped parameter models

Lumped parameter models (discrete models) are mathematical models consisting of a combination of masses, springs and dampers. The model parameters are determined by fitting the mechanical impedance of the model to an experimentally determined mechanical impedance, assuming linear system characteristics. The mechanical impedance is defined as the relation between a harmonic force applied to the head and a specific mechanical response of the head. If the frequency of the applied force is varied, both the amplitude and the phase of the response of the head change.

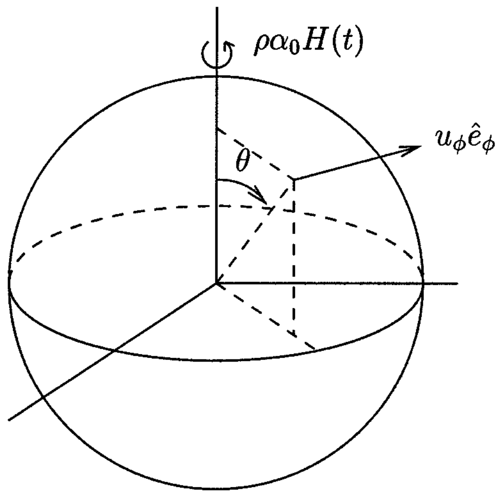


Figure 4.1: Spherical analytical model [Lee & Advani, 1970].

4.3.1 Translational models

Discrete models with one degree of freedom

Hodgson & Patrick [1968] showed that the response of a cadaver occipital bone to sinusoidal vibrations of the frontal bone can be modelled with a simple discrete model. The model consisted of a rigid mass connected in parallel with a linear spring and damper. The cadaver heads were filled with silicon gel and excited at frequencies varying between 8-1000 Hz (figure 4.2). Hodgson *et al.* [1967] had already shown that the results for cadaver heads filled with silicon gel were similar to the results of cadaver heads that contained intact embalmed brain. A remarkable amplitude decrease was seen only at the resonant frequency (313 Hz) as a result of the damping of the gel.

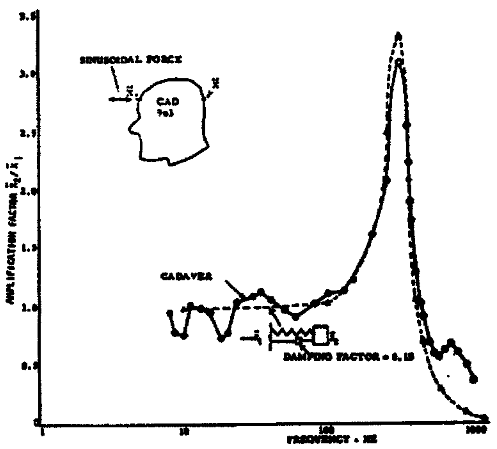


Figure 4.2: Occipital acceleration amplification for sinusoidal force input to the frontal bone of a cadaver (silicon filled cranial cavity), compared with a simple spring mass system having a natural frequency of 313 Hz and a damping factor of 0.15 [Hodgson & Patrick, 1968].

Following Hodgson & Patrick [1968], several other discrete head models were introduced. Slattenschek & Tauffkirchen [1970] presented a damped mass-spring model, called the Vienna Institute Model (figure 4.3), with $x(t)$ the relative displacement between the brain and the skull. This one degree of freedom model was based on the assumption that brain injury is caused by inertial loading of the brain.
The equation of motion for this model is:

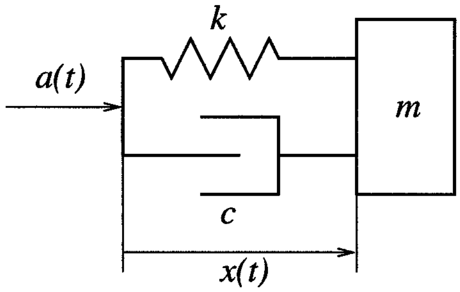


Figure 4.3: Vienna Institute Model [Slattenschek & Tauffkirchen, 1970].

$$\ddot{x} + 2\beta\omega\dot{x} + \omega^2 = a(t) \tag{4.1}$$

with

x, $\dot{x}$ and $\ddot{x}$ the relative displacement of the brain and its time derivatives;
$\beta = \frac{c}{2\sqrt{km}}$ the damping ratio;
$\omega = \sqrt{\frac{k}{m}}$ the natural angular frequency of the model;
a the acceleration pulse, measured at the head.

By using triangular acceleration pulses with average values taken from the WST-curve, Slattenschek and Tauffkirchen defined a tolerable skull-brain displacement $x_{tolr} = 2.35$ mm. The injury criterion set up is called the Vienna Institute Index (J) and the tolerance for this index is 1:

$$J = \frac{x_{max}}{x_{tolr}} < 1 \tag{4.2}$$

with

x_{max} the maximum relative displacement caused by an acceleration pulse;
x_{tolr} 2.35 mm.

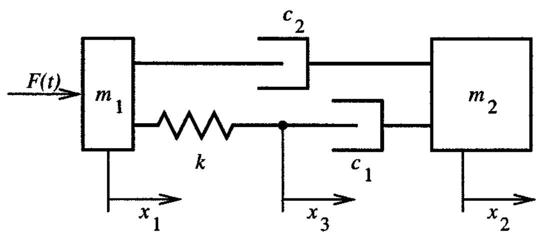
Revision of the Vienna Institute Index by further validations to the WSTC (especially concerning short duration pulses: 3-5 ms), resulted in the Effective Displacement Index (EDI) [Brinn & Staffeld, 1970]. A new maximum tolerable relative displacement was assessed: $x_{tolr} = 3.78$ mm.

Also based on the single mass system of the Vienna Institute Index is the Revised Brain Model (RBM) [Fan, 1971]. In order to be able to evaluate long duration pulses, both the displacement and velocity of the brain were determined with this model. Again a new value for x_{tolr} was determined (called $S_d = 31.5$ mm), which in this case is said to be valid for long duration pulses. For short duration pulses, the velocity of the brain (S_v) was considered a more important parameter with 3.44 m/s being the tolerance level. The injury criterion proposed thus becomes:

$\dot{x} < S_v$ for pulse durations < 20 ms;
$x < S_d$ for pulse durations > 20 ms.

Discrete models with two degrees of freedom

Using a two mass spring-damper system, Stalnaker *et al.* [1971] postulated a head injury criterion using mean strain, considered to be representative of brain deformation, as the parameter to predict head injury severity: the Mean Strain Criterion (MSC) [Stalnaker & McElhaney, 1970; Stalnaker *et al.*, 1971; McElhaney *et al.*, 1973]. The head model consisted of two masses, connected by one spring and one damper (parallel). Later studies indicated inconsistent behaviour of this model compared to cadaver responses and the model was changed by adding a second damper in series with the spring [Stalnaker *et al.*, 1985]. The new model is called the Translational Head Injury Model (THIM) and is shown in figure 4.4, together with its governing equations.



$$F = m_1\ddot{x}_1 + c_2(\dot{x}_1 - \dot{x}_2) + k(x_1 - x_3)$$
$$m_2\ddot{x}_2 = c_1(\dot{x}_3 - \dot{x}_2) + c_2(\dot{x}_1 - \dot{x}_2)$$
$$k(x_1 - x_3) = c_1(\dot{x}_3 - \dot{x}_2)$$

Figure 4.4: Translational Head Injury Model (THIM).

The THIM is used to develop a new head injury criterion, based on energy dissipated or stored by elements of the model. This new criterion is called the Translational Energy Criterion (TEC) [Stalnaker *et al.*, 1987; Rojanavanich & Stalnaker, 1989]. Mechanical impedance responses for various subjects (cadavers and animals) are determined experimentally [Stalnaker *et al.*, 1985, 1987; Rojanavanich & Stalnaker, 1989] and give different models for different subjects as well as for different directions of impact (A-P, P-A, S-I and L-R).

The physical meaning of the model elements (derived from cadaver experiments) is considered to be:

- summation of masses M_1 and M_2 will always add up to the total head mass;
- M_1 is the mass of the skull moving directly under rigid impact;
- stiffness K and damper C_1 form the non-linear skull stiffness in a given direction;
- damper C_2 is believed to be primarily the damping of the brain and found to be constant for all directions;
- the THIM concerns translational response to contact impact only;
- $F(t)$ is the impact force in rigid or padded head impacts.

Two types of head injury are considered with the THIM: brain concussion and skull fracture, which are primarily due to direct head impact. Analysis of the energies of the model elements as well as HIC values from the model, were found to correlate with head injury severity [Stalnaker *et al.*, 1987; Rojanavanich & Stalnaker, 1989]. The impact energy will go into dissipated energy from the two dampers and stored energy from the spring and the two masses. Regression analysis showed a good correlation between the energy dissipated in damper C_2 and the AIS scale, and close agreement between the power stored in spring K and skull fracture data was found.

4.3.2 Rotational models

The findings of Gennarelli [1981] and others, indicating the importance of rotational acceleration in the cause of head injury, incited Low & Stalnaker [1987] to develop a rotational discrete model of the human head. The aim of this model was to relate the rotational acceleration induced shear strains in the brain and its connective tissues to tolerance levels of the most severe head injuries, considered mainly to be caused by shear strains: SDH and DAI. Low and Stalnaker made the following assumptions and idealisations:

- the skull is rigid;
- the brain is a discrete system, symmetrical to the axis of rotation and with homogeneous, isotropic and linear material properties;
- a non-slip boundary condition in the skull-brain interface. The connecting tissues have viscoelastic properties, represented by groups of torsional Kelvin elements;
- the rotation of the brain is restricted to the sagittal plane.

The resulting head model was a system with two dimensions and three degrees of freedom as shown in figure 4.5. The brain was modelled with two masses with mass moments of inertia J_1 and J_2, connected by four torsional Kelvin elements. The contacting objects were modelled with a delayed step function, so that they would not add to the model's response until impact occurred. The physical characteristics of these contacting objects were given by the inclined springs and dampers K_d and C_d, connected to the reference frame. Both impact forces as well as distributed rotational acceleration were used as input for the model. It is however questionable whether these artificial models of the primate head resemble the human head in sufficient detail.

4.4 Finite element models

The advantage of FE models compared with for instance discrete or lumped parameter models is that the response of the different head structures can be evaluated locally. Field parameters such as pressures and shear stresses and strains can be visualised via contour plots and transient signals of local parameters can be visualised via x-y-plotting. When a FE head model has proven to simulate the dynamical brain response realistically, these local response parameters can provide very valuable information for getting a better understanding of brain injury mechanisms. Specific brain response parameters resulting from accident case simulations, such as for instance shear strain levels or local pressure maxima or minima could be found to correlate with brain injuries sustained during real accidents. This would make

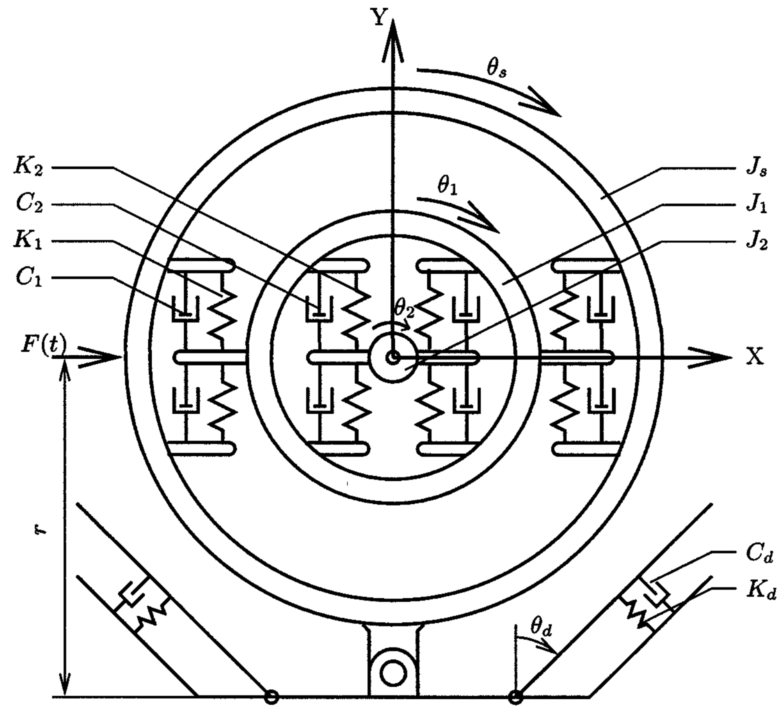


Figure 4.5: Rotational Head Injury Model (RHIM).

the FE head model a very valuable tool for predicting head injury risks. Furthermore, such correlations could give valuable indications on the nature of the mechanisms that result in different types of head injury.

4.4.1 Two dimensional finite element models

It was in the early seventies that finite element modelling entered the research field of head biomechanics. This section deals only with 2D finite element models. Most of them are plane strain parasagittal models, although some model the head in a coronal (or frontal) plane and whilst only few are transversal plane strain models.

Shugar [1977] published a 2D parasagittal FE model. The skull is represented as a closed rigid medium (figure 4.6). The brain, firmly connected to the skull, is an elastic, rather than viscoelastic, material and the membranes are not taken in account. The objective of this study was to analyse the influence of the cervico-occipital articulation on the intra cerebral stresses and strains under frontal and occipital impact.

Khalil & Hubbard [1977] proposed a parametric study of head response by FE modelling in the transversal plane. This study concentrated on determining the influence of the skull mechanical properties on the intra cerebral pressure field. The brain is modelled by an inviscid fluid and the skull by three different elastic models: an homogeneous shell, an oval shell and a three-layer spherical shell. The scalp was also modelled by an elastic material.

Theoretical load levels required to produce skull deformation and brain damage by cavitation were predicted. The results revealed that the load spatial distribution was the most important parameter in the head response. Another aspect studied by this team was the

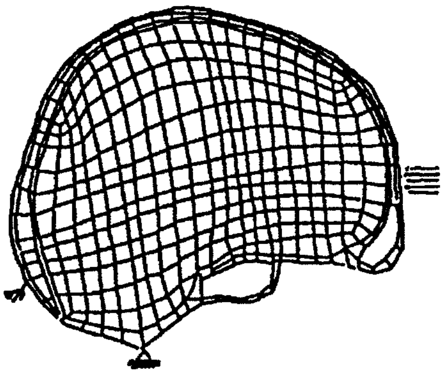


Figure 4.6: 2D parasagittal FE model [Shugar, 1977].

influence of the dimensions of the model on the intracranial pressure. They concluded that larger heads may tolerate slightly higher loads than smaller heads.

Others [e.g. Cheng *et al.*, 1990; Ruan *et al.*, 1991], conducted studies of head response to side impact using two dimensional coronal brain sections. Ruan *et al.* [1991] found that the coup and contrecoup pressure varied substantially and the natural frequency of the model increased from 49 Hz to 72 Hz when the membranes were simulated.

4.4.2 Three dimensional finite element models

As computer capabilities improved, more realistic material models were used and the head models were described with more anatomical detail. Although the development of 3D finite element models began approximately 20 years ago, many more models have been proposed in the last five years than in the previous 15. Ward & Thompson [1975] published one of the first 3D finite element models of the head. The model contained approximately 300 elements. It simulated the membranes and the foramen magnum. Natural frequencies of the head where found to be 23, 29, 32, 33 Hz.

The model was validated with the results of tests where static crush on cadavers was measured using X ray techniques. The low brain-skull relative displacement measured at the moment of impact suggested that the boundary condition allowed no movement between the brain and the skull. This hypothesis was later rejected by the neurosurgeons and neuropathologists.

One of the most recent FE head models is shown in figure 4.7. This model model consists of approximately 8,000 solid elements. This is a model with little anatomic detail, but with contact interfaces between the brain and the skull, representing the Cerebro Spinal Fluid.

An other recent model, developed by Claessens [1997] uses a different approach. It is described with more analytical detail (figure 4.8), but relative movement between the skull and the brain is not allowed. The model consists of approximately 12,000 elements.

Several others also contributed to the development of 3D Finite Element head modelling [eg. Ruan *et al.*, 1997; Zhou *et al.*, 1996; Kang *et al.*, 1997; Krabbel, 1997]. It is worth noting that all the models complement each other, each with its own unique contribution. However, the uniqueness of the models described, is in the validation process and even though the validation processes are often encouraging, important difficulties remain. Difficulties that

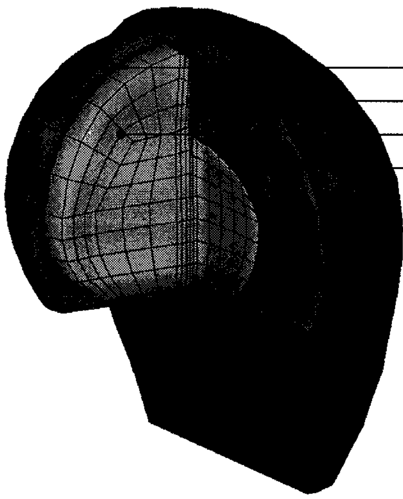


Figure 4.7: Example of a FE head model [Bandak & Eppinger, 1994].

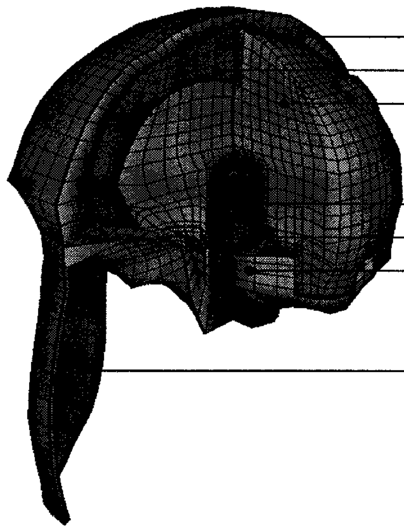


Figure 4.8: Example of a FE head model [Claessens, 1997].

arise in FEM modelling of the head are [Ward, 1983; Khalil & Viano, 1980; Ward, 1982]:

- representation of the geometry, which should at least include partitioning folds of dura and the foramen magnum
- modelling of the compressibility of the brain
- material properties (non-linear, anisotropic, inhomogeneous, solid-fluid interaction, etc.)
- validation of the model by comparing detailed model responses and overall cadaver or animal test data

- modelling of the skull (boundary conditions, face and scalp effects, variations in skull thickness)
- exact representation of the impact in terms of direction, magnitude and location
- modelling the interaction between the skull and the brain (so far little attempt has been made to model relative motion between these parts).

The main difficulty in modelling is in defining the equations for the brain material(s). Only two parameters were used to define a linear elastic material, Young's modulus E and the shear modulus G. The bulk modulus K and Poisson's ratio ν were then defined by the following equations:

$$G = \frac{E}{2(1+\nu)} \tag{4.3}$$

$$\text{and } K = \frac{E}{3(1-2\nu)} \tag{4.4}$$

Table 4.1 shows values for the shear modulus G and the bulk modulus K of the modelled brain tissue, used in several FE head models. In these brain tissue material models, G varies form 33 kPa to 228 kPa, and K varies from 20 GPa to about 5 MPa. The use in some models of a very low value for the bulk modulus can be explained because the modeller has assumed that the effect of the foramen magnum could be modelled by the compressibility of the brain. The eigenfrequencies (fn) are given in the final column, in case a modal analysis was carried out.

The rapid development of finite element codes has increasingly led to the use of linear viscoelastic material models. This material behaviour is characterised by the following equation:

$$G(t) = G_\infty + (G_0 - G_\infty)\mathrm{e}^{-\beta t} \tag{4.5}$$

where G_∞ is the long term shear modulus, G_0 is the short term shear

modulus and β is the decay constant. The values used in the bibliography for these three viscoelastic characteristics are given in table 4.1 as well and are approximately 50 kPa for G_0, 25 kPa for G_∞ and 100 s^{-1} for β.

Depending on the integration method, used for computing the stresses from the strains, numerical problems like *locking* or *hourglassing* can be expected [Bathe, 1996] for Poisson's ratios approaching 0.5. Although these problems can cause substantial errors, none of the researchers reported what countermeasures they used to prevent these errors. These numerical problems can be the cause of the big differences in the results (final column).

4.5 In conclusion

The lumped mass (discrete) models that have been reviewed have provided an insight into the simple behaviour of the head and the brain and have set useful precedents for the more detailed finite element models. However, all the discrete models have the same severe limitations: neither the location nor the severity of the head injuries can be predicted, the type of injury is only inferred from the measurements on the assumption that if certain values for the parameters are exceeded, then a certain range of injuries may be predicted.

Table 4.1: Mechanical characteristics of FE head models.

		Linear Elastic				Viscoelastic			Observation
	substr	K kPa	E kPa	G kPa	ν	G_∞ kPa	G_0 kPa	β s^{-1}	
Shugar 77	Brain	2.2E6							fn=43 Hz
	CSF								
Khalil 77	Brain	2.19E6							
	CSF								
Pluche 85	Brain	11.5E3	690	230	0.49				
	CSF	16E6	100	33	0.499999				
Cheng 90	Brain	1.6E3	100	33	0.49	16.2	49.0	145	
	CSF								
Ruan 91	Brain	2E5	66.7	22.5	0.48				fn=80 Hz
	CSF	11E3	66.7	22.2	0.499				
Chu 91	Brain	4.2E3	250	83	0.49				
	CSF								
Willinger 92	Brain	5E3	675	228	0.48				fn=120 Hz
	CSF								
Chu 94	Brain	4.2E3	250	83	0.49				ξ=0.5%
	CSF								fn=119 Hz
Ward 75	Brain	9.8E3	650	218	0.489				fn=50 Hz
	CSF	100E3	31.1033	10.103	0.45				
Hosey 82	Brain	11E3	66.7	22.2	0.499				
	CSF	11E3	66.7	22.2	0.499				
Dimasi 91	Brain					34.5	69.0	100	
	CSF								
Mendis 92	Brain								ν, E nonlinear
	CSF								
Ruan 92-94	Brain	2.2E6	5256	1680	0.4996				
	CSF	2.2E4	14520	500	0.489				
Bandak 94	Brain	1.8E6				34.5	69.0	100	cumulative
	CSF								damage
Krabbel 94	Brain								skull
	CSF								deformation
Willinger 95b	Brain	5E3	675	228	0.48				ξ=9%
	CSF	1.6E3	100	33	0.49				fn=100 Hz
Claessens 97	Brain	8.3E3	1000	338	0.48	169	338	50	fn=79.3 Hz
	CSF								

It has been shown that finite element modelling is the only method that can predict intracerebral parameters such as pressure, principal strains and stresses, as well as relative displacement of the principal head components. The variation in total skull thickness is well reproduced and 'sandwich' elements have been used to estimate skull deformation correctly. Different anatomical head characteristics such as the foramen magnum, the falx cerebri and the tentorium have also been incorporated in the more recent models. However, the quality of a numerical model is very much dependent on the material models and material parameters used. Thus, it is vital that the correct material parameters are used.

Chapter 5

Helmets

The fact that helmets can prevent and reduce head injuries is commonly accepted, but the exact manner in which they protect the head is not understood. Current helmets are empirically designed to meet the shock absorption requirements of a test standard. Several test standards are discussed in section 5.3. In most cases, the helmets are designed by experience and by trial-and-error methods.

5.1 Composition of helmets

Figure 5.1 illustrates the individual components of a full-face motorcycle helmet. The shell of the helmet consists of three different layers: the comfort padding liner, the protective padding liner and the outer shell. A chinstrap is used to prevent loosing the helmet prior to and during an impact.

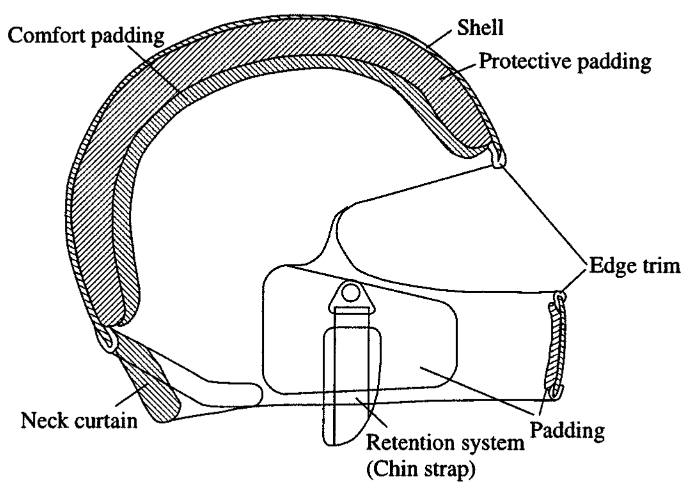


Figure 5.1: Cross section of a full-face motorcycle helmet [United Nations, 1994].

The purpose of the comfort padding liner is to increase the wearing comfort of the helmet and to provide a good fit on the head. It consists of low density, flexible, open-celled polyurethane or PVC foams and is often faced with a cloth layer [Gilchrist & Mills, 1993].

The protective padding liner, or impact liner, is the main impact energy absorbing component of a helmet. It is usually made of Expanded Polystyrene (EPS). The liner thickness applied in helmets is limited by aerodynamic design requirements and varies from about 25 mm to 40 mm. The mass density of the EPS foam applied in motorcycle helmets varies from approximately 30 to 90 kg/m^3.

The outer shell of a motorcycle helmet has three main functions: the distribution of the impact load over a large area, the prevention of penetration of sharp edged objects and the prevention of injury as a result of abrasion along a rough object. Two types of material are used in outer shells of motorcycle helmets: Rubber Reinforced Thermoplastic (RRT) or Thermosetting Fibre Reinforced Plastic (FRP)

5.2 Mathematical modelling of a helmet

One way of gaining more insight in the way helmets protect the head is by means of numerical simulation. Two types of numerical models for describing the impact behaviour of the helmeted head are found in literature: Lumped Mass Models and Finite Element Models (FE-models).

5.2.1 Lumped mass models

In lumped mass models, the (components of the) helmet and the head are modelled as rigid masses connected by massless springs and dampers. Mills & Gilchrist [1988] used such a model to simulate the helmet deformation as a result of impacts by flat and hemispherical strikers. They concluded that careful design of the softer foams inside the helmet may be of major importance in improving helmet performance. Gilchrist & Mills [1993] improved this model by including the effect of force oscillations of the shell mass on the elastic part of the liner (figure 5.2). They looked at the influence of impact velocity and performance in second impacts. They concluded that the thickness and stiffness of the comfort foam could improve the test performance of a prototype helmet. The scalp and hair of the average motorcyclist provide additional layers which may be more energy-absorbing than the comfort foam.

The lumped mass models are only capable of describing deformation for one specified loading condition. With these models, it is impossible to calculate the stiffnesses of the individual helmet parts from their shapes, dimensions and material properties.

5.2.2 Finite Element Models

FE models do not only allow the modelling of the mechanical properties of the helmet components, but also include the geometry of the helmet. This allows the influence of the interaction between helmet and head to be investigated. Also, there is only need for one model to investigate the effect of several different impact sites and rotational accelerations can be computed as well. The latter advantage is important because these accelerations may be of great influence to head injury.

Only a small number of finite element models of a helmet was found in literature [Köstner & Stöcker, 1987; Yettram *et al.*, 1994; Brands *et al.*, 1997]. None of them took into account the effect of the soft comfort liner, that provides the fit of the helmet on the head. The first two were not validated, but were used for trend studies only. Table 5.1 gives an overview of the three finite element helmet models.

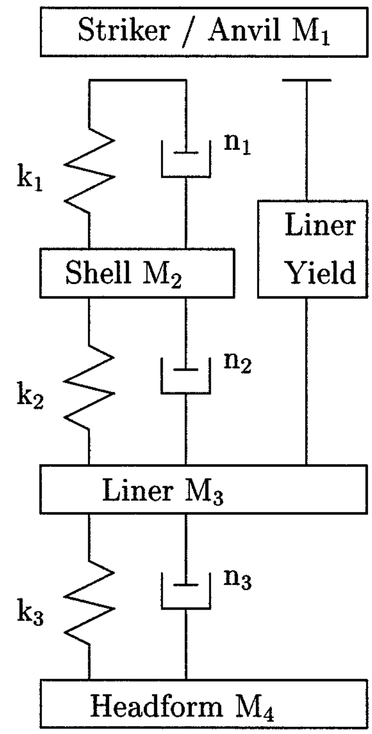


Figure 5.2: Improved lumped mass model of a helmet [Gilchrist & Mills, 1993].

Table 5.1: Comparison between finite element helmet models.

References	Helmet type	Materials				impact site
		shell	protective padding	comfort padding	head	
Brands *et al.* [1997]	full-face	ISOLIN	ISOPLA	gap	rigid	P, B, X
Köstner & Stöcker [1987]	full-face	ISOLIN	ISOPLA	-		B
Yettram *et al.* [1994]	full-face	ISOLIN	ISOPLA	-	solid	P

ISOLIN = isotropic, linear elastic; ISOPLA = isotropic, elastoplastic
P = top impact, B = frontal impact, X = rear impact

Köstner & Stöcker [1987] varied shell thickness, padding thickness, padding density and impact energy. Their simulations resulted in the following effects:

– increase in head acceleration with an increase in the shell thickness,

– increase in head acceleration with an increase in the padding density,

– minor reduction in acceleration with an increase in the padding thickness,

– maximum head acceleration is very dependent on the impact energy.

To reduce the effect of impact energy on head acceleration, their opinion was that paddings with an ideally plastic material behaviour should be used, which, combined with a rigid shell, should ensure constancy of acceleration over a wide range of impact energy. The use of materials with a honeycomb structure between shell and comfort padding may be a step in this direction.

Yettram *et al.* [1994] only varied shell stiffness (thickness 4 mm) and padding density (thickness 25 mm). They also used a 'theoretical material', a hypothetical material which would have an even lower density than the least dense of the actual materials. They computed peak accelerations and HIC values (table 5.2).

Table 5.2: HIC values and peak accelerations [g] (between parenthesis) for a variety of different shell stiffnesses and liner densities.

Shell		Liner			
Material	Young's Modulus [GPa]	Theoretical material	24 [kg/l]	44 [kg/l]	57 [kg/l]
GFRP	90.0		4722(425)	13247(690)	14360(709)
GFRP	70.0		4941(423)		
GFRP	40.0		5067(416)		
GFRP	20.0		4903(429)	10910(725)	11445(713)
PC	3.8		4659(369)	7480(520)	7780(535)
PC	2.1	3173(304)	4195(340)	6814(486)	7239(507)
HDPE	0.7	2833(291)	3366(321)		
LDPE	0.2	3069(352)	2977(328)		
No shell		1730(213)			

Although this model is not specific to any proprietary crash helmet, and thus the numerical values of peak acceleration and HIC should not be considered in relation to any values specified in relation to injury, the trends that emerge do correspond to those found in experimental investigations [e.g. Beusenberg & Happee, 1993; Hopes & Chinn, 1989]. These trends confirm that for optimal head protection, the shell should be more compliant and the liner less dense than is currently the case.

Brands *et al.* [1997] conducted a parametric study as well as a model validation using experimental data from drop tests (figure 5.3). The material parameters they used in their model are displayed in table 5.3.

Table 5.3: Material parameters of the helmet model used by Brands *et al.* [1997].

Component	Young's Modulus [MPa]	Poisson ratio [-]	Density [kg/m^3]	thickness [mm]	yield stress [MPa]
Protective padding	1.77	0.0	58.7	- -	0.32
Outer shell	$8.54 \cdot 10^3$	0.325	2082	3.0	- -

The results from their parametric study are given in table 5.4 (top impact). The explanation of a_{peak1}, a_{peak2} and a_{min} is given in figure 5.4. The outer shell does not contribute to an increase of impact energy absorption, because no hysteresis was modelled in this component. However, Mills & Gilchrist [1988] showed that in reality the outer shell shows hysteresis behaviour. Brands *et al.* [1997] also found that the material behaviour of the protective padding

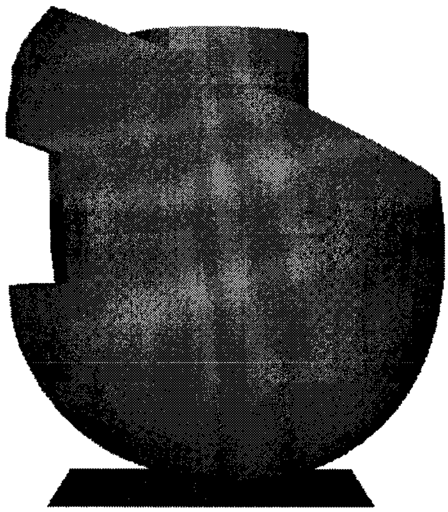


Figure 5.3: Finite Element Model of a full-face motorcycle helmet [Brands *et al.*, 1997].

Table 5.4: Results of the parametric study of Brands *et al.* [1997].

Parameter	a_{peak1}	a_{peak2}	$\frac{a_{peak1}-a_{min}}{a_{peak2}-a_{min}}$	Energy absorption
Protective padding				
$E_{\text{prot. padding}}$	+	+	+	+
$\rho_{\text{prot. padding}}$	o	o	o	o
$\sigma_{\text{yield, prot. padding}}$	++	++	+	-
Outer shell				
$E_{\text{outer shell}}$	o	o	-	o
$\rho_{\text{outer shell}}$	++	o	+	o
gap	-	-	+	o

Change of the observed quantity when parameter is increased from -30% to +30% of the original value: 0 less than 5%

-/+ decrease/increase in absolute value with more than 5%

- -/++ decrease/increase in absolute value with more than 15%

liner has a significant influence on the energy absorption of the helmet, however they were not able to use a correct material model for this component.

Brands *et al.* [1997] also validated their model for other impact sites than top impact. They found that the mass density of the protective padding liner should not be modelled constant over the entire helmet and that the comfort liner should be modelled in order to improve the helmet model.

Impact situations, often induce vibrations. Cappon [1997] performed a modal analysis on the outer shell of a full-face helmet to study its vibration modes. Since the helmet as a whole contains materials with considerable damping (for instance the protective padding liner) it seemed not appropriate to them to use modal analysis techniques on the complete helmet in

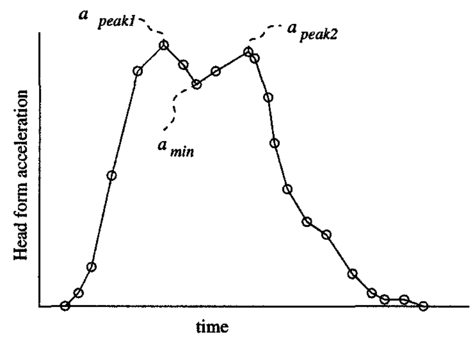


Figure 5.4: Typical head form acceleration pulse during droptest.

the first place, since modal analysis presupposes lightly damped systems.

They found that the magnitude of the Frequency Response Functions (FRFs) in the modal analysis was dependent on the impact force applied to the shell. As the impact force increased, the FRF-magnitude decreased and eigenfrequencies shifted to lower frequencies. This points toward the non-linear behaviour of the system. A unique relationship between impact level and FRF-magnitude had not been found, thus eliminating the possibility to extrapolate to higher impact levels as occurring in real accidents.

5.3 Review of helmet standards

Some very important and publicly available standards were chosen for a comparative study. The reviewed standards are given in appendix A. These standards are the major standards in Japan (JIS T 8133), Australia (AS 1698), Great-Britain (BS 6658), the USA (ANSI Z90.1) and the United Nations (ECE R22-04). Also, standards from two major motorcycle safety organisations (Federal Motor Vehicle Safety Standard 218 and Snell Memorial Foundation Standard) and an international ISO standard (R1511) were studied. It should be noted however that these standards apply to transport only and do not cover sport or leisure activities.

Table 5.5 shows which standards require a test for which quality. This list applies only to those features relating to head protection. Investigations for visor or flammability tests are not included.

Resistance to penetration The ability to resist an impact with objects which cause localised loads such that the head will be penetrated.

Shock absorption The object of a requirement for shock absorption is to reduce the risk of a head/brain injury from impact forces. Shock absorption can be assessed from accelerometer outputs or by measuring the energy absorbed.

Rigidity Is the ability of a helmet to withstand compressive loads and is determined from quasi-statically applied forces.

Measurement of friction This can be divided into two parts:

- loads generated by abrasion
- loads generated by contact with protrusions

Table 5.5: Standards and tests.

Standards	Penetration	Shock	Rigidity	Friction	Retention
ECE R22-04-1995		X	X		X
ISO R1511-1970		X	X		X
Snell 1995		X	X		X
BS 6658:1985		X	X		X
JIS T 8133-1982		X	X		X
AS 1698-1988		X	X		X
ANSI Z90.1-1992		X	X		X
FMVSS 218-1988		X	X		X

Both will tend to impart rotational motion to the head.

Retention system effectiveness Examines the helmet retention system's ability to withstand external loads and to hold the helmet on the head during an accident.

This comparative study of standards concentrates on the two most important safety features designed to reduce the potential for head and neck injury: resistance to penetration and shock absorption capacity.

5.3.1 Penetration test

The main consideration is the choice of test velocity and hence test energy of the impactor, that best represents an impact between the head and object during a real accident. Table 5.6 summarises the details of the specifications in the different regulations.

Table 5.6: The resistance to penetration required by helmet standards.

Standards	Test method	Drop mass [kg]	Drop height [m]	Criteria x_{min} [mm]	Impact energy [J]	Retention system
ECE R22-1991	A	3	1	>5	29.4	relaxed
ISO R1511-1970	A	3	1	>5	29.4	adjusted
Snell 1990	B	3	3	>0	88.3	relaxed
BS 6658:1985	B	3	2	>0	58.9	fixed
JIS T 8133-1982	B	3	1	>0	29.4	relaxed
AS 1698-1988	B	3	3	>0	88.3	relaxed
ANSI Z90.1-1992	B	3	3	>0	88.3	relaxed
FMVSS 218-1988	B	3	3	>0	88.3	adjusted

Test Method There are two principal test methods and these are designated A and B in table 5.6. In method A (figure 5.5), a metal punch with a conical head is placed onto the top of the helmet which is mounted on a fixed test head. A metal drop hammer of 3 kg mass is then impacted onto the top of the punch and the depth penetrated is measured by an external instrument, such as a photoelectric device.

In method B (figure 5.5) the helmet is mounted on a fixed test head and impacted by a sharp object, also with conical head form rounded at the top. The penetration is assessed from marks on the head form.

The angle of the cone forming the conical head (methods A and B): 60°
Radius of rounded top of conical heads is a maximum of 0.5 mm.

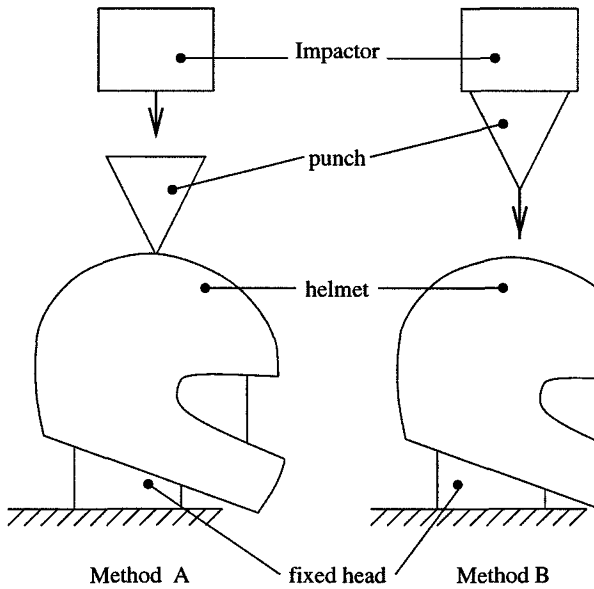


Figure 5.5: Two principal penetration test methods.

Criteria Method A: the distance between the striking object and the head form shall not be less than x_{min} during the test process and this is measured using an external optoelectric device.

Method B: The head form is examined visually for possible contact and the helmet will fail if contact has occurred.

Impact Energy The impact energy depends on the drop mass and drop height, and varies between 29.4 J and 88.3 J. The impact energy should be related to what occurs in accidents, the different standards use very different impact energies. Although in real accidents the circumstances are also divergent, this divergence is not visible within a penetration standard, therefore it is not clear why the difference between standards is so large.

Generally penetration injuries are very infrequent and do not cause brain injuries, therefore this sector attracts less attention from standard-makers and indeed ECE Regulation 22-04 does not require a penetration test any more.

The two test methods are very similar and there is no substantial difference between them. The only question is that of repeatability. With method A, the positioning is easier and it is possible to test over a greater surface area of the helmet. However, the impact energy is more clearly defined for method B.

5.3.2 Impact (shock absorption) test

This requirement is assessed from the results of a helmet drop test. The primary objective of an impact energy absorption requirement is to assess the ability of a helmet to protect a rider by measuring parameters such as, maximum acceleration in the head form, HIC and possibly other criteria which can be related to the potential for injury. However, the severity of the test is a function of the impact energy and the anvil profile and this varies between standards. Table 5.7 compares the test conditions and requirements of different standards.

Table 5.7: Shock absorption requirements of helmet standards.

Standards	Anvil Shape	Drop height [m]		Cumulative	Peak	Method
		1^{st} Impact	2^{nd} Impact	duration [g]	[g]	
ECE R22-04-1995	flat	2.50	non	150/5ms	300	B
	hemisphere	non	1.83	150/5ms	300	
ISO R1511-1970	fixed helmet	dropping striker (5kg)		300/5ms	400	A
Snell 1990	flat	2.35(2)*	1.72(2)*	- -	300	C
	hemisphere	2.35(2)*	1.72(2)*	- -	300	
	edge	2.35(2)	non	- -	300	
BS 6658:1985	flat	2.87	1.43	- -	300	C
	hemisphere	2.5	1.27	- -	300	
	flat	2.15	1.08	- -	300	
	hemisphere	1.83	0.94	- -	300	
JIS T 8133-1982	flat	1.83	1.83	150/4ms	300	C
	hemisphere	1.38	1.38	150/4ms	300	
AS 1698-1988	flat	1.83	1.83	200/3ms	300	C
	hemisphere	1.38	1.38	150/6ms	300	
ANSI Z90.1-1992	flat	2.43	1.84	- -	300	C
	hemisphere	2.43	1.84	- -	300	
FMVSS 218-1988	flat	1.83	1.83	200/2ms	400	C
	hemisphere	1.38	1.38	150/4ms	400	

(2)* - 2 impacts at each site

The main parameters used for evaluating helmet impact energy absorption are:

- resultant acceleration or vertical acceleration versus vertical load
- impact velocity (or drop height) and impact energy
- form of anvil.

Method A specifies a free falling striker impacting onto the helmeted head form fixed on a base.
Method B specifies a guided free falling helmeted head form impacting onto different anvils.
Method C specifies a guided free falling helmeted head form attached to a support arm impacting onto different anvils.

The force is measured only for the ISO standard (method A) and it shall not exceed $F_{max} = 19.6$kN. The resultant acceleration is measured in method B and vertical acceleration is obtained in method C. The permitted maximum acceleration varies between 300g and 400g. Some standards specify an average acceleration which must not be exceeded over a specified duration of time. This is shown under "cumulative duration" in table 5.7.

Three different impacting shapes (anvils) are specified: flat, hemispherical and edge (kerbstone). Head form mass varies from 3 kg to 6 kg. Head forms are made from magnesium alloy, except for the ISO head form which is made from layers of hardwood.

These standards measure and calculate only the linear accelerations. Rotational effects should also be taken into consideration in a new helmet standard, because rotational acceleration is the most important cause for severe head injury: SDH and shearing injury [Gennarelli, 1981; Ommaya, 1988].

Although the majority of helmets sold in Europe are full-face helmets, in fact there are very few definitive requirements for chin guards within the current standards, even though a great part of the injuries is caused by chin impact. Only the SNELL-Standard (used in America) [1985] and the British Standard Institution [1985] actually detail tests that should be carried out on chin guards.

5.4 In conclusion

Lumped mass models have been very useful in parametric studies, to investigate the influence of the different components on the dynamical behaviour of a helmet. However, these models are only valid for one impact direction and are only capable of computing translational accelerations.

To investigate the way in which the helmet protects the head, it is necessary to use 3D finite element modelling. But, as with the human head models it is vital that the correct material model is used, especially since the protective padding liner has a complex material behaviour.

One significant difference between the work of Brands *et al.* [1997] and the work of Köstner & Stöcker [1987] and Yettram *et al.* [1994], is the influence of the density of the protective padding. Because of the structure of the material, the yield stress and Young's Modulus are dependent on the density of the material. In the parametric study by Brands *et al.* [1997], these parameters were varied independently.

That current helmets afford good protection is in no doubt, but it is clear that there is much room for improvement. Efficient energy absorption with a minimum tendency to induce rotational motion and a comprehensive evaluation of the whole helmet including the chin guard of a full-face helmet are features which require special attention.

Currently, only the British Standard 6658 includes tests for rotation and the chin guard, and only Regulation 22-04 requires an assessment against HIC (time dependent criterion). Although ECE Regulation 22-04 (recently amended from -03) is widely used, it does not require tests for rotation or the chin guard.

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Appendix A

List of standards examined

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