DEVELOPMENT AND ASSESSMENT OF A SURFACE FORCE ABDOMINAL SENSOR

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ABSTRACT

Abdominal injuries occur seldom but are often of high severity. Various proposals for the assessment of the abdominal injury risk have been made. Nevertheless only a few dummies are equipped with abdominal sensors.
With support from the European CHILD and APROSYS projects abdominal surface force sensors for Q child dummies and the Hybrid III 50 percentile dummy were developed. The surface matrix force sensors are able to assess the time history of the applied force and the location of the load, which is important as different abdominal regions are meant to require different load limits. The sensors for the Q dummies were used in the CHILD project reconstruction programme. The analysis of 14 accident reconstructions indicates a very good correlation between the applied load assessed by the proposed sensor and the AIS 3+ injury risk. However, the number of cases is still small. Additional reconstruction cases should be able to validate the described results.

INTRODUCTION

Abdominal injuries do not occur very frequently but when they do occur they are often very severe. Although measures against submarining, which was the main cause for abdominal injuries in the past, were established (e.g., seat ramps, belt pretensioners and improved belt geometry), recent accident investigations by LAB (common research institute of Peugeot and Renault) show that there are still abdominal injuries in real world accidents, especially in rear seat occupants [Walfisch, 2002]. Based on the study of US accidents with air bag deployment, Digges et al. [Digges, 1996] proposed the measurement of abdominal injury risk to be a matter of priority. Within the combined work programme of EEVC WG 12 (Crash Dummies) and 18 (Child Safety) the abdomen was defined as an important body region for children using booster seats [Jager, 2005]. However, abdominal injuries were also observed in children using a seat with integral harness.
The current dummies offer various possibilities to measure injury related loads. To assess abdominal injuries, various criteria and possibilities have been proposed. Nevertheless, only a few of them are used in current adult dummies (e.g., EuroSID, THOR). Furthermore, there are currently no appropriate sensors being used in child dummies to measure abdominal injury related loads, even though children have a considerably higher abdominal injury risk compared to adults.

ANATOMIC BACKGROUND

Abdominal organs are either, thin-walled and hollow (stomach, intestine, urinal bladder etc.) or sponge-like and blood-filled (liver, kidneys, spleen etc.), these are the so-called solid organs. It is important to know that solid organs are located in the upper abdomen, while the hollow ones are generally located in the lower abdomen. However, hollow organs can be found in the upper abdomen as well (e.g., stomach). Hollow and solid organs behave totally different under mechanical loads. Subgroups to be considered in the field of abdominal injuries are children and pregnant women. The latter subject is not discussed in this paper.
There are some differences between adults and children to be considered. One of the main differences is that children's ribs and musculature provide less protection of the organs. For example, the liver is almost completely covered by the ribs in adults, but is only partially covered in children. In addition to this, the abdomen of a child is bigger in relation to height than that of an adult. The function of the organs is the same for children as it is for adults.

ACCIDENT STATISTICS

Based on GDV (association of the German insurance institutes) accident investigations, frontal accidents between 1993 and 2000 with air bag equipped cars Roselt et al. [Roselt, 2002] showed that the abdomen is the second priority for critical injuries (AIS 4/5). In the analysed sample cars with model years up to the year 2000 were analysed. 60% of the sample included models between the years1996 and 1998, and 80% between 1995 and 1999. This means that more recent models of cars were included in this sample. Figure 1 shows that injury frequency decreases significantly with injury severity for most of the regarded body regions.

However, the abdomen shows more AIS 4 injuries than AIS 3 injuries.

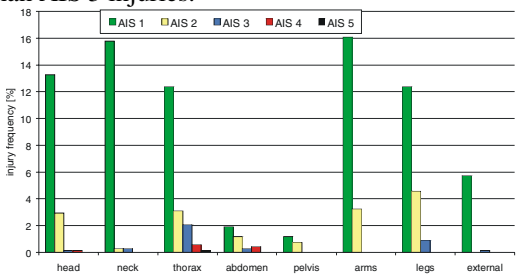

Figure 1. Injury frequency and severity in airbag equipped cars (288 driver) [Roselt, 2002].

The analysis of the AIS 3+ injuries shows that the thorax is most often affected by severe injuries, while legs and abdomen are more or less at an equal level at the second rank. A comparable situation can be observed when the weighted injury frequency (harm) is considered (Figure 2). The abdomen is at an equal level with arms and head, while the thorax is the most affected body region followed by the legs.

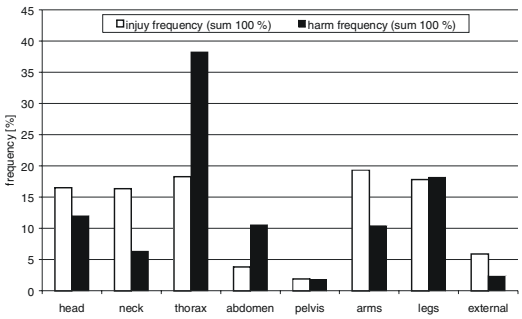

Figure 2. Injury frequency and harm in airbag equipped cars (288 driver) [Roselt, 2002].

Comparison of the drivers and passengers with AIS 3+ injuries shows that the injury severity for passengers is generally lower than for the drivers (Figure 3).

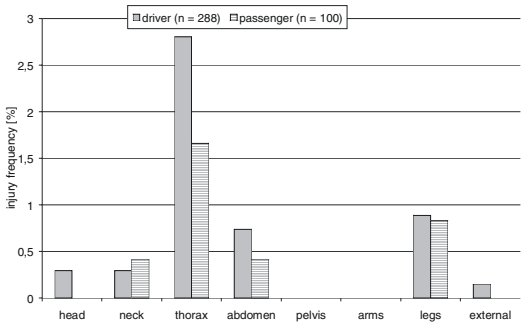

Figure 3. Comparison of AIS 3+ injuries for drivers and passengers [Roselt, 2002].

An earlier study in the UK based on accidents between the years 1984 and 1986 shows that 11% of the occupants sitting on the struck side of the car received abdominal injuries of which 52% were AIS 3+ rated [Harms, 1987]. The contacts causing abdominal injuries in lateral impacts are doors, door furniture and, in the more severe accidents, intruding objects. In frontal collisions, 20% of the restrained occupants receiving AIS 3+ injuries were hurt at abdomen or lower back. However, it is not possible to divide these injuries into two separate categories; abdominal injuries and lower back injuries. Newer data from the UK and Germany are presented below in the section concerning comparison of airbag-equipped cars with cars without airbag.
To discuss the influence of airbags on the injury distribution amongst body regions, the next chapter compares accidents with and without airbags.

Comparison with/without Airbag

To investigate the differences between airbag equipped and non-airbag equipped cars, Frampton et al. [Frampton, 2000] analysed UK accident data from the year 1992 to 2000 and German accident data from the year 1996 to 1999. The sample included front seat occupants in frontal crashes only. It can be seen, that abdominal injuries occur slightly more often in airbag equipped cars than in those without airbags (Figure 4). Because of the significant reduction of head and spine injuries in the airbag cases, the relative injury outcome of abdominal injuries is now higher for cars equipped with airbags than for those without.

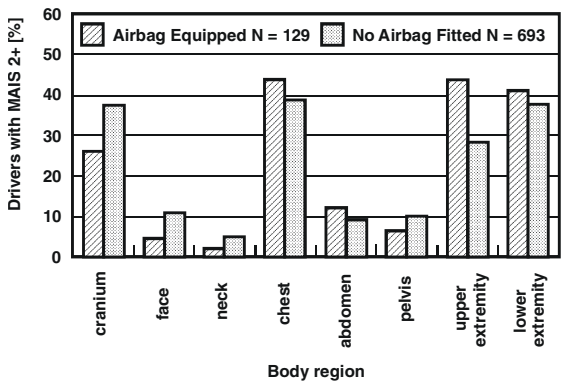

Figure 4. AIS 2+ body region rates for belted drivers with MAIS 2+ (UK) [Frampton, 2000].

Children

Children must be regarded separately because of their biomechanical differences regarding pelvis and abdomen.
Based on GDV data of German accidents during 1990 and 1991 [Langwieder, 1997] it is obvious that children using a CRS properly received the majority of their severe injuries to the head, neck, chest and abdomen (Figure 5).

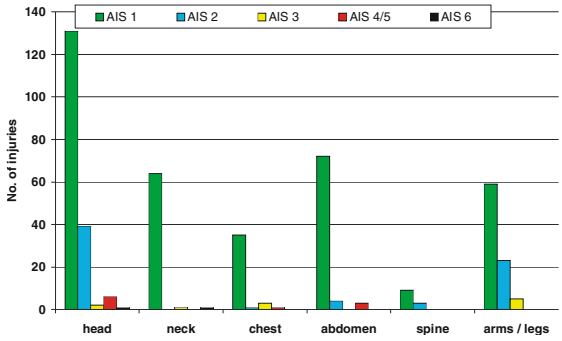

Figure 5. Injuries of 415 children [Langwieder, 1997].

Regarding only those children restrained with a CRS (harness type), there is almost the same distribution of injuries (Figure 6).

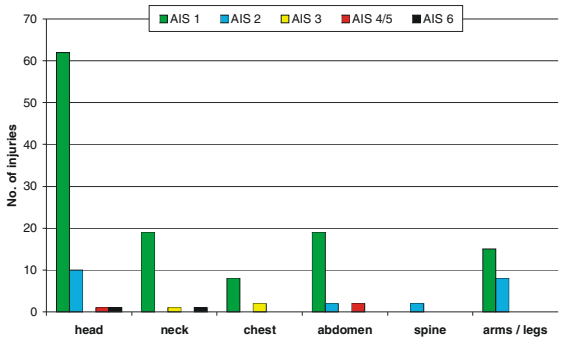

Figure 6. Injuries of 200 children using a CRS [Langwieder, 1997].

Taking into account the "harm" of the different body regions, the abdomen is again shown to be the second ranking region followed by the extremities, neck and chest (see Figure 7).

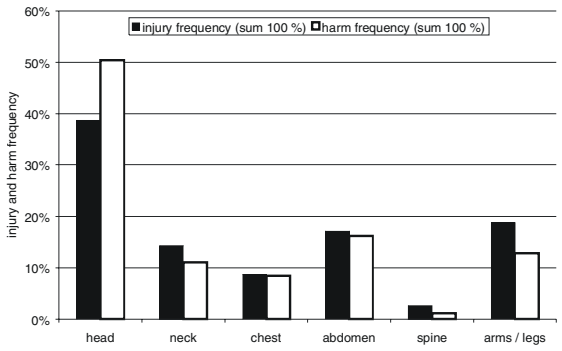

Figure 7. Calculated "harm" for different body regions (415 children) [Langwieder, 1997].

An older study based on the same accident data of the years 1990 and 1991 shows that wearing a 3-point belt or lap belt only was the main problem resulting in abdominal injuries to children [Langwieder, 1994]. However, even children using a CRS suffer abdominal injuries, see Table 1.

Table 1.
Number of injuries to different body regions depending on restraint system [Langwieder, 1994]

	Number of injuries AIS 1 – 6		
	CRS	3 point only	Lap only
Head	34	26	4
Neck	13	11	1
Chest	6	6	-
Abdomen	12	17	5
Arms	5	9	1
Legs	5	5	2
T/L Spine	-	2	1
Total	75	76	14

Based on French accidents of the years 1992/1993 and 1995/1996, Trosseille et al. [Trosseille, 1997] came to the conclusion that abdominal injuries occur mainly in older children, with an age above three years, using either boosters or car belt without any CRS.

INJURY PATTERN

Most of the abdominal injuries are located at the skin, the musculature and organs. While the severity of skin and musculature injuries is mostly minor, the severity of organ injuries can be serious. The main injuries to solid organs are crushing or bursting of the organ, laceration and rupture. Hollow organs sustain contusion, haematoma, perforation and laceration, as well as rupture of organs. Besides the risk of bacterial contamination of the abdominal cavity with the intestine content, rupture of organs leads to massive bleeding.
The most affected abdominal organs are liver, spleen and kidneys. All of these are solid organs, which have a higher injury risk [Cavanaugh, 1986]. The injuries are caused mainly by contact with lap belt, steering wheel, armrests, etc.

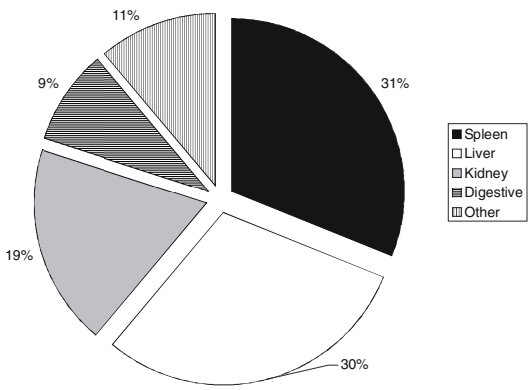

Figure 8. Distribution of the injured abdominal organs for the years 1993 – 1998 [Yoganandan, 2000].

The distribution of injuries amongst the abdominal organs was, for example, investigated by Yoganandan et al. [Yoganandan, 2000], based on NASS

data of front occupants in all accidents except rollover of the years 1993 to 1998. Based on this study, the liver and spleen are the most common abdominal organs involved, followed by the kidneys and the digestive system (Figure 8).
The same study shows that abdominal injuries are most common in belt restrained occupants, which is true for frontal, lateral and oblique impacts [Yoganandan, 2000].
The distribution of abdominal injuries for different accident types shows almost an equal share of frontal and lateral impacts (Figure 9).

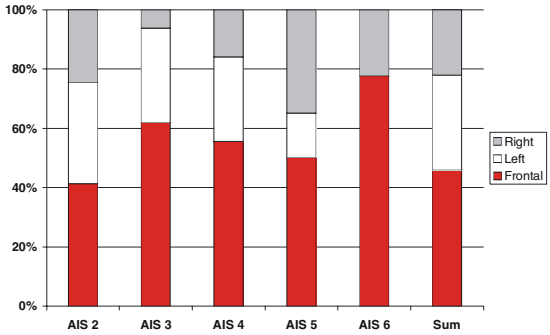

Figure 9. Abdominal injuries dependent on impact direction [Yoganandan, 2000].

In the following part, injury patterns for frontal impacts are described in more detail based mainly on case studies and biomechanical tests.
The main causes for abdominal injuries are the belt system for belted occupants and the steering wheel for unbelted occupants. However, because of the increased amount of equipment in cars with airbags, the number of steering wheel induced injuries is decreasing. On the other hand, very little data of abdominal injuries in airbag-equipped cars exists in the literature. Therefore, it is difficult to assess the injury pattern in these cars.
According to accident investigations in the UK, abdominal injuries in frontal car accidents between 1984 and 1986 are caused mainly by steering wheel and belt webbing (see Figure 10) [Harms, 1987].

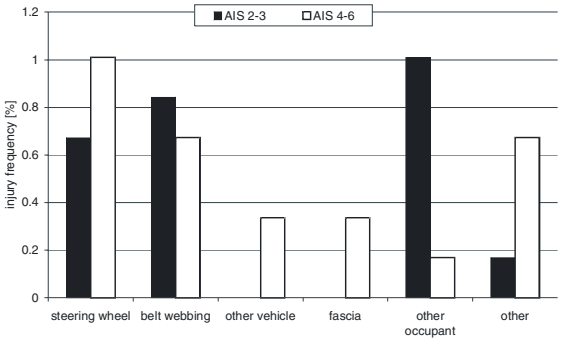

Figure 10. Contact location for abdominal injuries for 594 restrained front occupants [Harms, 1987].

Because of the different loading conditions the belt, steering wheel and airbag induced injuries are described separately.

Belt Induced Injuries

After the introduction of lap belts in cars a new injury pattern was observed. The abdominal injuries induced by the belt were called seat-belt injuries. Under this heading injuries to the abdomen and lumber spine were summarised. Due to improvements of the belt geometry and the introduction of the automatic three-point belts the situation has changed during the last decades.
Abdominal injuries in frontal impacts are mainly caused by the deceleration of the abdominal viscera (here tears, wounds or rupture of the viscera can be observed), flexion of the trunk around the pelvis (which causes pressure injuries by compression between thighs and trunk) and submarining [Leung, 1982].
The most important problem of belt induced abdominal injuries is the phenomenon known as submarining. Submarining describes the situation when the lap belt intrudes the abdomen. There are differences in submarining. In the normal case of submarining, the lap belt rides above the iliac crest during the accident. Another kind of submarining is when the lap belt is positioned above the iliac crest prior to the accident. In submarining cases injuries to liver, spleen, kidney and digestive system are reported [States, 1987]. Although submarining was mainly reported in the more distant past, there are still submarining cases seen in today's accident statistics; e.g., reported by Walfisch [Walfisch, 2002] for rear seat occupants.
However, seatbelt induced injuries seem to be possible without submarining. For example, Witte [Witte, 1968] reported 5 cases of abdominal injuries without submarining. Intestinal injuries seem to be possible because of acceleration of the digestive content against the seatbelt. The acceleration of the digestive content leads to longitudinal ruptures of the intestine, while the acceleration of the intestine itself leads to laceration of the attachments.
The risk of submarining is higher for the rear seat because of inclined knees and better-restrained torso. The higher risk for the rear seat was confirmed by an accident analysis in the UK from 1992 to 1995 [Cuerden, 1997].
One serious reported problem is misuse when the shoulder belt is worn under the arm. This is intended to increase comfort but decreases the efficiency of the belt system.
In Heidelberg, in cadaver tests, reported by Schmidt et al. [Schmidt, 1974], abdominal injuries were either caused by the shoulder belt (liver ruptures, spleen ruptures and kidney contusion) or by the lap belt (fat tissue and muscle tissue ruptures, mesentery ruptures and intestinal ruptures).

However, Leung et al. [Leung, 1982] reported cadaver tests in which only a few injuries were caused by the shoulder belt.
Cadaver tests conducted by APR showed 47 out of 70 cases received abdominal injuries [Leung, 1982]. Only 23 of these 47 cases with abdominal injuries were submarining cases. For the non-submarining cases, no intestine, colon or mesentery injuries were reported. Liver injuries were on an equally low level for submarining cases and non-submarining cases. Based on this study, 68% of the abdominal injuries are caused by the lap belt.
In a review of the US NASS data from 1988 to 1994 Elhagediab et al. [Elhagediab, 1998] it was found that about 17% of the abdominal injuries were belt induced. These injuries are mainly to the digestive system (Figure 11). In the data set, frontal collisions with un-ejected occupants were investigated.

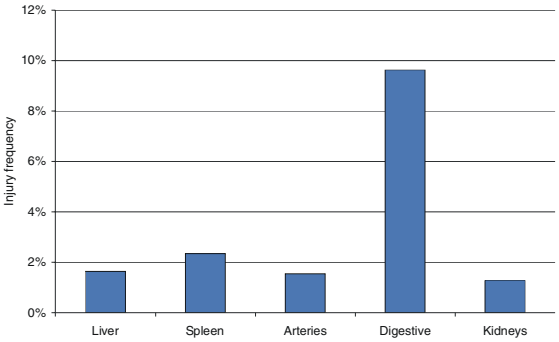

Figure 11. Injury frequency for belt induced injuries [Elhagediab, 1998].

Children need to be regarded separately, because of the lap belt positioning problems and the reduced protection provided for the abdominal organs. Based on French accident data from 1992/1993 and 1995/1996, abdominal injuries to children only occur in older children (age above 3 years) restrained with a booster, or without any CRS, using only the vehicle's belt [Trosseille, 1997]. The affected organs were mainly liver, spleen and intestine. Generally, children restrained by a 3-point belt and booster were injured at the abdomen because of the lap belt position [Walfisch, 2002].

Steering Wheel

Abdominal injuries induced by the steering wheel are mainly a problem for unbelted drivers. This is the reason why there are considerable differences between US and Europe. These differences can be found in accident statistics and is also reflected in research programmes.
Based on the review of the US NASS data from 1988 to 1994 Elhagediab et al. [Elhagediab, 1998] mentioned above, it was found that about 69% of the abdominal injuries are steering wheel induced. These injuries are mainly to the liver (Figure 12).
In the data set, frontal collisions with un-ejected occupants were investigated.

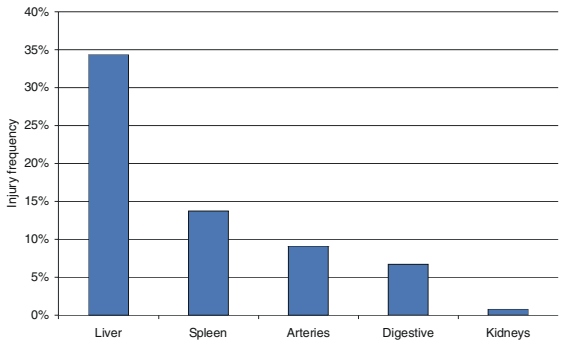

Figure 12. Steering wheel induced injury frequencies [Elhagediab, 1998].

Liver injuries, due to contact with the steering wheel, are mainly laceration of the liver or laceration of the central venous junction between the liver lobes [Lau, 1987].

Airbag

As previously explained, the number of accidents with airbags is too small to assess the injury mechanism in a reliable manner, but trends can be observed.
Augenstein et al. [Augenstein, 1996] investigated frontal accidents with airbag deployment. The most severe injuries occurred in drivers (restrained by 3-point belt, shoulder belt only or without belt) at chest and abdomen. Abdominal injuries were observed in all restraint conditions. They were rated between AIS 2 and AIS 5. Most of the abdominal injuries in this sample were reported for the liver, but spleen, kidneys, pancreas and intestine were also injured. Liver injuries were mainly to the front lobe in severe crashes and to the rear lobe in less severe accidents.

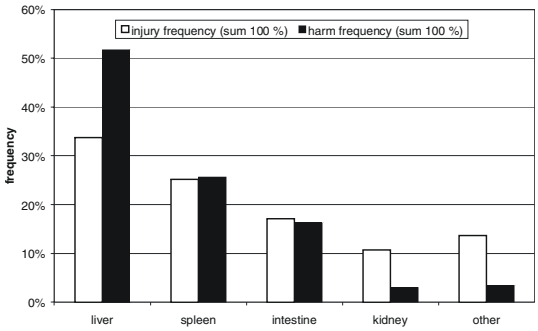

Figure 13. Harm distribution within abdominal region [Augenstein, 1998].

In a later study, Augenstein et al. [Augenstein, 1998] came to the conclusion that about 30% of the injury weighted harm is related to abdominal injuries. Again the liver is the most affected abdominal organ, see Figure 13.

Based on the above-mentioned review of the US NASS data of the years 1988 to 1994, Elhagediab et al. [Elhagediab, 1998] reported that airbag induced abdominal injuries are quite rare and only spleen injuries are associated with the airbag. Dishinger et al. [Dishinger, 1996] came to a different result concerning airbag induced abdominal injuries. Although most abdominal injuries could be reduced by an airbag the number of kidney injuries increased, Figure 14. These findings were again based on data from US accidents collected by the Maryland Hospital in 1993 and 1994. However, accidents with airbag deployment are normally of higher severity than those without (if the car is equipped with an airbag). Accident severity is not considered in this sample.

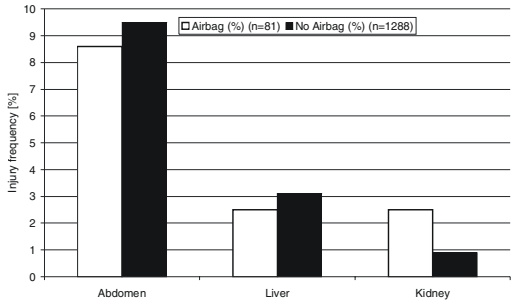

Figure 14. Comparison of selected injuries depending on airbag deployment (belted drivers) [Dishinger, 1996].

Synthesis of Injury Pattern

Most of the analysed data are either old or are coming from the US. Car occupants in the US are often not restrained by belts. For the data outside the US, the restraint system changed over the years – modern cars are equipped with airbags, three point belts with pretensioner and load limiter. Abdominal injuries could be caused by the belt system, steering wheel and airbag. With respect to steering wheel induced injuries, differences between US and European data need to be considered, because the steering wheel impact location is mainly only relevant for unbelted drivers.

Taking into account the injuries caused by the belt system, most of the reported cases are submarining cases. Due to the improvements of the belt system this kind of injury mechanism will decrease with modern cars. However, for rear seat passengers it is still relevant because of the worse belt geometry and lower equipment rates with pretensioner and load limiter. Besides the submarining cases abdominal injuries were reported to be caused by the shoulder belt or the lap belt without submarining. Steering wheel induced abdominal injuries are seldom for belted occupants. Due to improved equipment rates of cars with airbags they are likely to disappear completely.

There are indications that airbags could induce abdominal injuries as well. Due to the low number of airbag-equipped cars included in the accident databases, it is not yet possible to prove this situation.

INJURY CRITERIA

Three mechanisms causing abdominal injuries are possible: blunt trauma, penetration and acceleration. Injuries caused by acceleration are generally ruptures of organs and blood vessels. Bones (e.g., broken ribs) and vehicle parts can cause penetrations. Because of the improvement of the vehicle interior, penetration of abdomen is of less importance.

Impact location, direction and magnitude have a significant influence on injury severity due to mobility, location and natural protection of the abdominal organs.

Most of the abdominal injuries observed after road accidents are caused by blunt trauma. For more detailed answers concerning an appropriate injury criterion a lot of tests with surrogates (e.g., human cadavers, living anaesthetised and cadaver animals) have been conducted. But mostly it remains unclear whether force, intra-organic pressure, compression of abdomen, velocity or a combination of these correlate well with the injury severity. In fact, there is a correlation between force, pressure and compression. The impact speed seems to have an important influence on injury severity. In addition to that, the stiffness of the abdominal organs and, therefore, the stiffness of the abdomen itself is dependant on the impact velocity.

Besides dummy abdominal measurements, other criteria were proposed. For example the EEVC subgroup on Biomechanics came to the conclusion that the risk of injuries to the lower part of the abdomen could be reduced by lap belt positioning criteria. In addition, hard parts of the belt (e.g., the buckle) should not come into contact with the abdominal wall. Upper abdominal injuries were thought to be covered by thorax protection criteria [Halpern-Herla, 1976].

In addition, it could be necessary to have different injury criteria for localised loads (e.g., by belts) and distributed loads (e.g., by airbags) [Elhagediab, 1998].

Within the following chapters the findings explaining a selection of possible injury mechanisms (e.g., compression, force, impact speed) will be described. It is important to consider the test conditions. The pre-test body posture and the preparation of test subjects have an important influence on the abdominal behaviour and the possibilities to detect abdominal injuries. The type of organ (hollow or solid) and impact direction are also important. In addition, the available measurement techniques are of importance.

Compression

Going back in history, compression was the first proposed abdominal injury criterion. From today's point of view it worked well as long as the compression rate was more or less at the same level (e.g., for belted occupants). In the context of abdominal injuries, compression means the relative abdominal deflection for frontal impacts and the relative half width abdominal deflection in lateral impacts.

This chapter summarises findings which do and do not support compression as an appropriate criterion.

In fixed back tests of anaesthetised canine subjects a correlation of intestine compression and the occurrence of abdominal injuries was found by Wiliams [Wiliams, 1966].

Experiments with anaesthetised primates and human cadavers showed that abdominal compression was related to injury severity [Stalnacker, 1973].

Miller [Miller, 1989] found a correlation between maximum compression and injury severity in tests with anaesthetised porcine subjects loaded with a safety belt.

In steering wheel tests with anaesthetised swine compression was found to correlate "also" good to injury severity [Miller, 1991]. But in the tests with impact velocities between 1.7 and 12.4 m/s VC_{max} (see below) it was found to correlate better.

A "quite good" correlation of compression to injury severity (correlation factor 0.64) was found in frontal steering wheel impacts to anaesthetised porcine subjects [Lau, 1987]. The viscous criterion was found to correlate better.

Investigation of hepatic injuries with unembalmed human cadavers showed that tolerance levels based on velocity, compression or combination of both seemed to be inappropriate [Nusholtz, 1985]. But it has to be considered that several tests were applied to the same cadaver.

In tests with anaesthetised swine subjects, Lau et al. [Lau, 1988] found that the compression was inappropriate. The probability analysis of liver laceration risk shows a poor result, Figure 15.

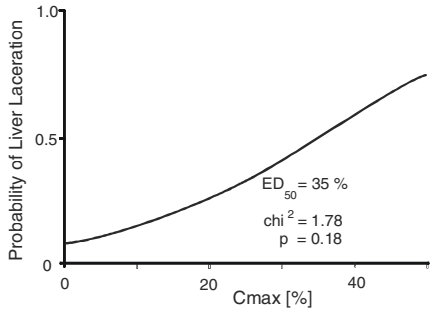

Figure 15. Analysis of liver laceration [Lau, 1988].

Lateral impactor tests to the right side of anaesthetised New Zealand White Rabbits showed better correlation of compression for liver injuries than for kidney injuries [Rouhana, 1986].

In lateral impact tests to exposed human livers and PMHS with different velocities and compressions, deflection was found to be a better parameter than compression. Deflection is measured in cm and not normalised. For the tests with an impactor disc with a diameter of 15 cm, the correlation of deflection to injury severity was R = 0.85 [Talantikite, 1993].

Walfisch et al. [Walfisch, 1980] found no obvious relation between compression and injury severity in lateral drop tests of unembalmed human cadavers; this was explained to be caused by the pre-impact body posture. The drop height was 1 and 2 m, which correlates to 4.4 m/s and 6.2 m/s respectively. However, there were no injuries observed for compression below 28% and compression was found to be more easily measured with a dummy. Therefore, a switch which activates at compression above 28%, was proposed for the EuroSID.

The main criticism on compression as an injury criterion is that the abdominal viscera behaves rate sensitive. This means that the abdomen is able to deform without any injury when the blunt loading is applied slowly [Penberthy, 1952]. Crushing of organs is possible with high anteroposterior compression (40 – 60%) and low speeds while fast loading (> 12 m/s) can lead to severe injury with low compression (10 – 20%) [Lau, 1981].

Compression was found to be a good predictor with low impact velocities (< 3 m/s, e.g., for belted occupants) but in sports and for unbelted occupants or belted occupants in high speed accidents compression is not appropriate [Viano, 1988].

Viscous Criterion (VC)

The viscous criterion was initially proposed for thoracic injuries. The product of compression (C) and velocity (V) takes into account the rate sensitive behaviour of the abdominal organs. Dependent on the measurement capabilities of the different authors, either the product of the maximum velocity (normally the initial velocity) and the maximum compression ($V_{max}C_{max}$) or the maximum of the continuously calculated product of compression and velocity were regarded as VC. In some publications, modifications to the pure VC criterion were proposed.

Rouhana et al. [Rouhana, 1984 and Rouhana, 1985] found a good correlation of injury severity and $V_{max}C_{max}$ in lateral impact tests with anaesthetised rabbits. Based on these experiments, the Bounded Abdominal Injury Criterion (v*C /(1-C)) was proposed.

Stalnaker et al. [Stalnaker, 1985] confirmed the relevance of $V_{max}C_{max}$ for the prediction of injuries in subhuman primates.
In steering wheel tests with anaesthetised swine, VC_{max} was found to correlate best with injury severity [Miller, 1991]. The impact velocity was varied between 1.7 and 12.4 m/s in these tests.
A good correlation of VC_{max} with injury severity was found in frontal steering wheel impacts to anaesthetised porcine subjects [Lau, 1987] (Figure 16). Within these tests, only liver injuries were observed.

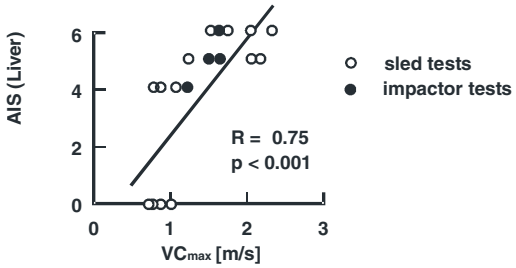

Figure 16. Correlation between abdominal injury severity and VC_{max} [Lau, 1987].

Tests with different velocities and a defined compression of 16% with anaesthetised rabbits showed good correlation of injury severity with $V_{max}C_{max}$ [Lau, 1981]. The test speed varied between 5 and 20 m/s.
Investigation of hepatic injuries with unembalmed human cadavers showed that tolerance levels based on velocity, compression or a combination of both (VC_{max}) seemed to be inappropriate [Nusholtz, 1985]. But it has to be considered that several tests were applied to the same cadaver. Therefore, the allocation of injuries to the impact conditions does not seem to be clear.
In tests with anaesthetised swine subjects, Lau et al. [Lau, 1988] found good correlation of VC_{max} and injury severity in frontal, lateral and oblique impacts.
Comparison of tests with liver laceration and those without showed clear differences between the maximum viscous criteria (VC_{max}) assessed in the tests, Figure 17.

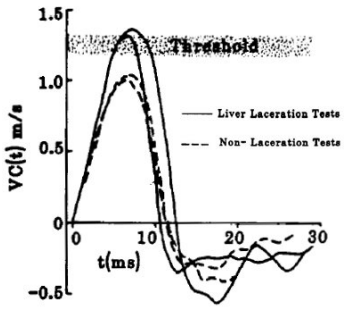

Figure 17. Tests with minor liver laceration versus no injury [Lau, 1988].

In right side lateral impactor tests to anaesthetised New Zealand White Rabbits, Rouhana et al. found good correlation of $V_{max}C_{max}$ with injury severity [Rouhana, 1986].
Lateral pendulum tests with unembalmed human cadavers to chest, abdomen or pelvis showed the best correlation for VC_{max} with abdominal injury severity [Viano, 1989]. The tests were conducted to cadavers in an upright position. It has to be considered that several tests were conducted with one cadaver. Therefore, the allocation of injuries to impact condition seems to be problematic although different body regions were impacted.
Based on frontal steering wheel impact tests with anaesthetised swine Viano et al. [Viano, 1988] proposed VC_{max} as an appropriate injury criterion.
Impactor tests with a 15 cm diameter disc impactor to exposed fresh human livers and fresh human cadavers showed a correlation coefficient of VC_{max} with injury severity of 0.71 [Talantikite, 1993]. In the PMHS tests, the impactor hit the side of the body approximately at the COG of the human liver. Comparing the measurement results of VC_{max} and force each showed a correlation coefficient of 0.78.

Force

Most of the published studies are focussing on peak force and do not regard the time history of the load. Force was often found not to be a useful indicator of abdominal injuries because the measurement of abdominal forces was considered to be too complicated [Rouhana, 1987 and Walfisch, 1980].
In impact tests to different anaesthetised primates and porcine subjects, impactor force and duration correlated to injury severity [Trollope, 1973]. The tests were conducted with different impactor properties, of which the impactor size influenced the injury severity (smaller impactor causes higher injury severity). In addition, the subject weight had an important influence.
Stalnaker [Stalnaker, 1973] found a correlation of abdominal injury severity and the logarithm of impact force and time duration squared.
Peak force showed best correlation with AIS 4+ injuries of all biomechanical measurements in rigid lateral pendulum tests to the mid abdomen of cadavers [Viano, 1989].
Miller [Miller, 1989] showed that peak force is well correlated with AIS 3+ injuries and AIS 4+ injuries in belt loading experiments.
In steering wheel tests with anaesthetised swine, force was found to correlate “also” well to injury severity [Miller, 1991]. But in the tests with impact velocities between 1.7 and 12.4 m/s, VC_{max} was found to correlate better.
Comparison of tests with pregnant baboons showed that the belt tension was the only indicator, which differentiates between tests with and without injuries [Snyder, 1966].

Investigation of submarining tests with human cadavers showed that injuries were observed when the belt tension after submarining exceeded a certain limit [Leung, 1982].
In right side lateral impactor tests to anaesthetised New Zealand White Rabbits a correlation of impact force and $V_{max}C_{max}$ to injury severity was found [Rouhana, 1986]. Dependent on the boundary conditions, advantages were found for one or the other.
In lateral impact tests to exposed human livers and PMHS with different velocities and compressions, maximum force was found to "strongly correlate" to the impact velocity. For the tests with an impactor disc with a diameter of 15 cm the correlation of maximum force to injury severity was R = 0.73 [Talantikite, 1993]. The deflection of the abdomen also correlates with the impact force. However, the correlation of force and velocity is better.
In lateral drop tests of unembalmed human cadavers Walfisch et al. [Walfisch, 1980] found that the normalised force is a reliable indicator for injury severity (R = 0.98).

Comparison of Different Criteria

The outcome of the different studies, concerning abdominal injury criteria described above, are summarised in Table 2. Within this table "+" means that the author felt that the criterion is a good predictor of the abdominal injury risk, while "-" stands for the opposite. When a criterion was not investigated the cell is left blank.
Nusholtz et al. [Nusholtz, 1994] compared the different injury criteria in impactor tests with unembalmed, pressurised, sitting, human cadavers. In these tests, a part of a steering wheel rim contacted with the abdomen at the height of the 2nd lumbar vertebra. The comparison covers compression, deflection, energy loss, velocity, force and spinal acceleration. Most of the different measurements are dependent on each other (within 10% tolerance), except the spinal acceleration. This seems to be the main reason why the review of proposed injury criteria shows a blurred picture with different proposals for the injury criteria, depending on the type of test and the measured quantities in these tests.
One good source for the comparison of different criteria is tests conducted by Hardy et al. [Hardy, 2001]. These tests comprise human cadaver tests using seatbelt impactors, rigid bar impactors and airbag deployment. Compression, impact velocity and impactor force were measured for most of the tests. The main goal of this study was the assessment of force-deflection characteristics of the abdominal region. For the comparison within this study, only subjects that were impacted once were considered. Due to the different injury sensitivity of the different abdominal regions tested in this study, a comparison of different injury criteria based on these tests is not possible. However, a comparison of these tests does allow for analysis to see if the criteria are robust against other load conditions than originally defined for. Because of the available data, the $V_{max}C_{max}$ criterion has to be analysed. Concerning the belt tests, mainly injuries of the rib cage were reported. This includes rib fracture up to rib two. This is the reason why the seat belt tests were not considered within this study.

Table 2.
Summary of different results concerning abdominal injury criteria in frontal impact conditions

Source	Test subject	Test type	C	VC	F	Remark
[Wiliams, 1966]	Anaesthetised canine	Frontal impactor tests	+			
[Trollope, 1973]	Anaesthetised primates and swine	Frontal impactor tests			+	F and duration
[Lau, 1981]	Anaesthetised rabbits	Frontal impact tests with different V and constant C		+		
[Stalnaker, 1985]	Anaesthetised primates	Frontal impactor tests		+		
[Miller, 1991]	Anaesthetised swine	Frontal steering wheel tests (1.7 < v < 12.4 m/s	"also good"	+	"also good"	
[Lau, 1987]	Anaesthetised swine	Frontal steering wheel impact	"quite good"	+		
[Nusholtz, 1985]	PMHS	Frontal steering wheel impact	-	-		Several tests to one subject
[Viano, 1988]	Anaesthetised swine	Frontal steering wheel impact		+		
[Miller, 1989]	Anaesthetised swine	Frontal belt loading	+		+	
[Lau, 1988]	Anaesthetised swine	Frontal and oblique	-	+		
[Snyder, 1966]	Pregnant anaesthetised baboons	Frontal sled tests			+	Belt force
[Leung, 1982]	PMHS	Frontal sled, submarining			+	Belt force after submarining

The abdominal AIS was coded based on the injury description published by Hardy et al. [Hardy, 2001], the $V_{max}C_{max}$ value was calculated from the impact velocity and maximum compression. It is clear that the abdominal injury severity depends on the location of the load and the Hardy tests were conducted at different abdominal levels (mid abdomen, lower abdomen). Nevertheless, the injury severity of these tests is compared with the measured load, the abdominal compression and the $V_{max}C_{max}$ criterion in Figure 18.

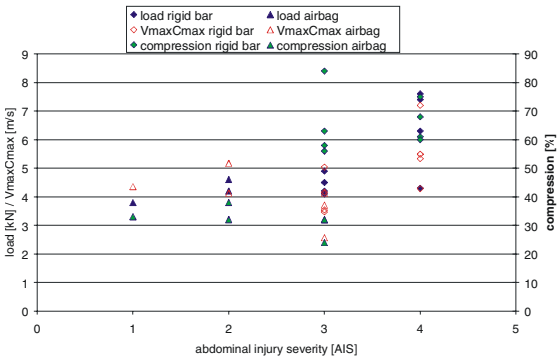

Figure 18. Re-evaluation of Hardy tests (data [Hardy, 2001]).

This comparison shows that in the airbag tests higher VC and lower compression is necessary to cause injuries of a certain level, while the force criterion seems to be independent from the load case. Analysis of the AIS 3+ injury risk of the tests leads to the same conclusion, Figure 19.

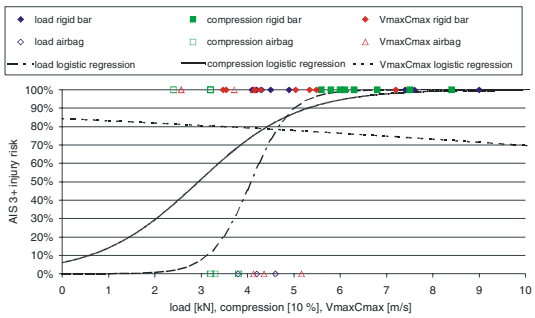

Figure 19. Re-evaluation of AIS3+ injury risk of Hardy tests (data [Hardy, 2001]).

Based on tests of Lau et al. [Lau, 1981], Talantikite et al. [Talantikite, 1993] and Nusholtz et al. [Nusholtz, 1994] it can be noticed that the impact force and VC are more or less linearly dependent on each other.
Another issue to be considered is the application range for which the criterion is applicable. Lau et al. [Lau, 1986] described the compression as valid for compression rates up to 2 m/s and the VC_{max} criterion to be valid in the range between 2 m/s and 30 m/s, see Figure 20.

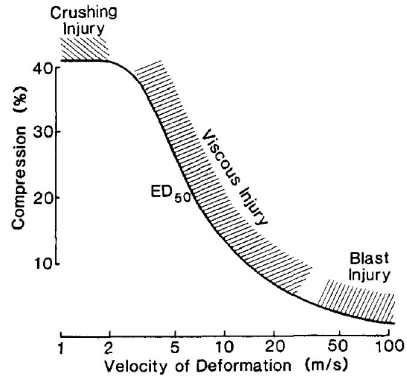

Figure 20. Range of validity for the viscous criterion and compression criterion [Lau, 1986].

While compression rates of 30 m/s and more are unusual in automotive accidents, low compression rates (submarining) are still possible.
From the theoretical point of view, airbag induced loads show a high compression rate combined with a distributed low compression. This combination seems to be problematic for a reliable VC measurement as small measurement mistakes of the compression measurement lead to considerable mistakes when computing the velocity (derivation of the compression).
In conclusion, it can be noticed that all three criteria can be considered as more or less equivalent. However, compression and VC do not seem to be valid for all kinds of load conditions for the abdomen (slow submarining, fast steering wheel and distributed airbag loading). Regarding the surface force, no constraints are known.

SENSOR DEVELOPMENT

The sensor development is based on a detailed state of the art review of proposed and currently realised sensor concepts; see [Johannsen, 2006]. These concepts are assessed based on the following criteria.

Requirements for Abdominal Sensors

The abdominal sensor should be able to cope with the following criteria and properties:

- Measurement of appropriate injury criteria (either contact force or VC)
- Time history measurement
- Detection of location of load
- Ability to assess loads applied by lap and shoulder belt, steering wheel (for adult dummies only) and airbag
- Reliable measurement in the sense of robust sensor and repeatable measurements
- Applicable for existing dummies without major changes to the dummy

- No significant influence on dummy behaviour
- Low purchase and operation costs (easy calibration and long calibration intervals)
- Usable with normal crash equipment (no additional data acquisition system or computing of the signals necessary)

The requirements mentioned above are a maximum design goal. For the first phase of the development, which was focussing on existing technologies, it seemed to be impossible to achieve all requirements. Therefore it was necessary to find a suitable compromise, which should take into account future development possibilities.

Sensor Description

The standard FlexiForce® A102 sensor is a thin (0.1 mm), flexible printed circuit. It is 14 mm wide and 203 mm in length (standard version). The active sensing area is a 9.5 mm diameter circle at the end of the sensor. The sensors are constructed of two layers of substrate, i.e., a polyester film. On each layer, a conductive material (silver) is applied, followed by a layer of pressure-sensitive ink. Adhesive is then used to laminate the two layers of substrate together to form the sensor.
The FlexiForce single element sensor acts as a resistor in an electrical circuit. When the sensor is not loaded, its resistance is approximately 20 MΩ. When a force is applied to the sensor, this resistance decreases. For the use in crashes, it is possible to connect the sensors to a Wheatstone´s Bridge by operating the sensor parallel to one of the resistors.
For the use as abdominal sensor, a matrix of a number of single sensors has to be designed taking into account a compromise concerning requirements for maximum distance between the sensors and the number of channels.

Abdominal Sensor for Q-Dummies

This chapter describes the Q-dummy family with emphasis on the abdominal region first and then the final abdominal sensors for the Q3 and Q6, which represent a 3 years old child and a 6 years old child, respectively.
The accident statistics shows that there is no risk for abdominal injuries for children using a rear facing CRS and a very low risk for children using a seat with integral 5-point-harness. Four-point-harness seats, which caused abdominal injuries in the past, are not available any more. Therefore, it is reasonable to limit the sensor development at the first stage to the Q3 and Q6, which represent children, which can be restrained by booster seats.
The sensors were developed within the CHILD project framework. They were used for the CHILD accident reconstruction programme.

Q-Dummy Family - The next generation of European child dummies – the Q family – offers the following dummies: Q0 (newborn), Q1, Q1.5, Q3, Q6 (the numbers are meant to give the age of the child that the dummies represent). The dummies are designed for multidirectional use, which requires abdominal sensors able to cope with at least frontal and lateral impacts. The modular design of the dummies allows using the same philosophy of the abdominal sensors for different dummy sizes.
Except for the Q0, the design of the "extended abdominal region" of all Q-dummies is composed of a rigid thoracic spine, which fixes the rib cage, houses a chest accelerometer and a chest deflection measurement device (capable of measuring either in X or in Y direction). An elastomer lumbar spine is mounted at the lower end of the thoracic spine, which connects the thorax to the pelvis. Between lumbar spine and pelvis a load cell can be installed. In addition, the pelvis houses another accelerometer. An abdominal block made of PU foam and covered with skin simulates the abdomen itself. The dummy is clothed with a wet suit.

Sensor Design - The chosen dummies are usually restrained by the vehicle belt with or without a booster seat or with a CRS with integral belt system. In addition to the belts, abdominal injuries can be caused by parts of the seat – in lateral and oblique impacts – and by the passenger airbag. The minimum width of CRS-belts is 30 mm, the average between 35 and 40 mm. Therefore, the distance between two "neighbour" sensors must be smaller than this width. In addition, it is necessary that the sensors be spaced equally to ensure a good coverage and efficient evaluation of the results.
For both dummies, 20 sensors are arranged in an array across the surface of the abdominal block. The array complies with the requirements mentionned above and the geometry of the abdominal surface.
The dummy design allows placing the sensors directly at the surface of the abdominal block. This allows assessing the loads applied to the "soft" part of the abdominal region. The chest deflection measurement device can assess the loads applied to the "hard" part of the thorax.
The following Figure 21 shows the sensor array for the Q3 dummy. The distance between two sensor centres is 35 mm; because of the diagonal arrangement of the sensors, the vertical and the horizontal distance are below 30 mm each.

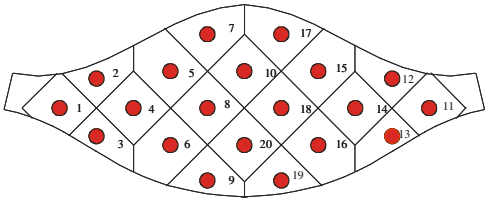

Figure 21. Sensor matrix for the Q3 dummy (front view).

Due to the different shape of the abdominal block for the Q6 the sensor arrangement is slightly different to that of the Q3. Except for four sensors, the distance between two neighbour sensor centres is 30 mm because of the diagonal allocation of the sensors that means a horizontal and a vertical distance of 25 mm.

Abdominal Sensor for the HIII 50th Percentile Dummy

This chapter summarises the Hybrid III design with emphasis on the abdominal region and the relevant load conditions as well as the sensor design for that dummy.

Description of the Hybrid III 50th Percentile Dummy - The Hybrid III dummies are designed for frontal, frontal oblique and frontal off-set collisions. Besides the 50th percentile dummy, a 5th percentile female and a 95th percentile male version of the HIII adult dummies exist.
The dummy is built of a metal skeleton and foam flesh material. It can house various sensors at head, neck, chest, pelvis and legs, consisting of mainly accelerometers and load cells. Regarding the "extended abdominal region" – from chest to pelvis – the dummy has a metal thoracic spine, which houses an accelerometer. Six metal ribs compose the rib cage. A sternum connects the ribs. A chest compression measurement device is fixed between spine and sternum. Soft tissue is simulated by a chest jacket, which offers different thicknesses at different locations of the chest. The thoracic spine is connected to the metal pelvis by a rubber lumbar spine. The pelvis is covered by foam material. An abdominal insert fills the abdominal cavity between chest and pelvis.

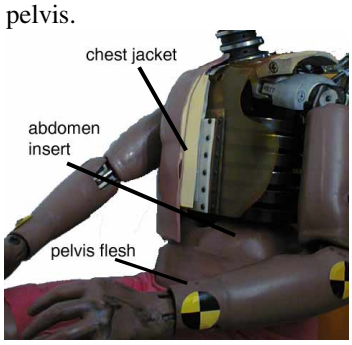

Figure 22. Relevance of the chest jacket for the abdominal response.

Although the chest jacket covers the abdominal region, it does not seem to have any abdominal function, see Figure 22. The jacket is quite thin below the dummy's rib cage. In the upper region, there is additional foam between jacket and ribs and the jacket itself is thicker. The pelvis flesh also does not seem to perform any abdominal supporting function.

Relevant Load Conditions for the Abdominal Sensor - Abdominal injuries for the adult driver in frontal accidents are induced by lap and shoulder belt, steering wheel and airbag – either separately or by a combination of them. The steering wheel is, of course, not relevant for the front seat passenger; while at the rear seat, lap belt and shoulder belt are the only source of abdominal injuries in frontal accidents today. Frontal accidents include frontal oblique and frontal off-set collisions. Although every new car is equipped with a driver air bag, steering wheel loading is applicable for reconstruction cases with older cars and multiple collision accidents, even with newer cars.
While loads applied by steering wheel or belt are local, airbags apply distributed loads to either the entire abdomen or large parts of it. Steering wheels have a diameter of about 380 mm – the rims of steering wheels have a diameter of approximately 30 mm. Standard belts have a width of 45 mm, but one has to take into account that the belt material bends along the longitudinal direction when the belt is tightened.

Sensor Description - Based on the loading cases mentioned above (lap belt, shoulder belt, steering wheel and airbag) the distance between two sensors should not exceed 25 mm in Y and in Z direction. Recognising this distance would prevent missing the load applied by a steering wheel with a rim diameter of 30 mm. As mentioned above, steering wheel loading is not valid for new cars in single collision accidents (as simulated by compulsory crash tests). If this load case is considered not to be necessary, the distance of two sensors should not exceed 35 mm for the belt loading cases. This distance would show enough overlap with the belt width, even when the belt is bent.
The sensors are to be applied at the anterior surface of the abdominal insert. This is the only part that covers the entire abdominal cavity. The chest jacket does not influence the abdominal function and the pelvis flesh does not reach the upper end of the dummy abdomen. For the assessment of the injury risk of the "hard abdomen" covered completely by ribs for the HIII dummy, it is reasonable to measure the rib compression. Although the localised compression measurement for the abdomen is felt to be a disadvantage it should not be a problem for the "hard thorax". The disadvantages are problems in localised measurement of compression of soft

material, if the load is applied at a different location than the localised measurement device is situated. The following Figure 23 shows the sensor matrix for the Hybrid III 50th dummy in a two-dimensional view. For this dummy, 26 sensors are used; in comparison to 20 sensors for Q3 and Q6 dummy. The horizontal and the vertical distance between two sensors is 22.5 mm.

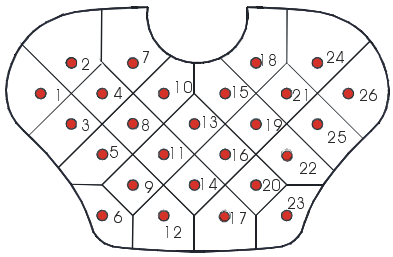

Figure 23. Sensor matrix for HIII 50th percentile.

ASSESSMENT OF THE ABDOMINAL SENSOR

The sensor concept has been assessed by a feasibility study, analysing the sensor behaviour under different load conditions and in the accident reconstruction programme of the EC funded CHILD project.
Amongst other criteria the average surface pressure was analysed regarding the correlation of the abdominal injury severity and the measured load.
The probability to sustain an injury of a certain severity level depending on the abdominal average surface pressure was analysed using the logistic regression method. Using this analysis method, one is looking for a clear shift in the injury risk. This clear shift allows the definition of a load limit at the location of the shift.
Figure 24 shows the probability for the AIS 2+, AIS 3+ and AIS 4+ injury risk in relation to the measured abdominal average surface pressure. While this injury criterion seems to be appropriate to predict the AIS 2+ and AIS 3+ injury risk, it is not for the prediction of the AIS 4+ risk.

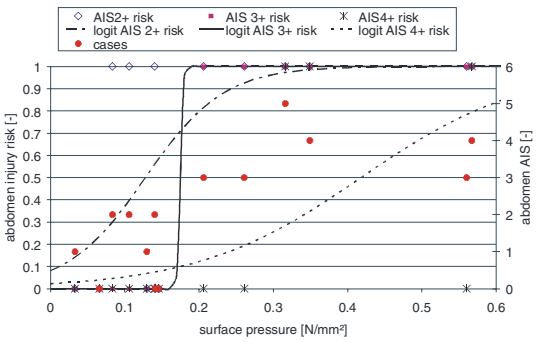

Figure 24. Probability analysis of abdominal injury risk depending on the average surface pressure.

The poor predictability of the AIS 4+ risk is likely to be caused by one of the fourteen analysed cases, where a very high load corresponds to AIS 3 injuries. Going into more detail, it becomes obvious that there is only a little information available in this case. The described intraperitoneal bleeding normally occurs secondary to an organ trauma. It is likely that more information could justify an injury severity coding of AIS 4.
The injury risk curves shown above indicate a load limit for 50% probability of AIS 2+ injuries of 0.13 N/mm² and for 50% probability of AIS 3+ injuries of 0.175 N/mm².
However, the chosen FlexiForce sensor shows considerable disadvantages with respect to reliability. Sensor sensitivity is not stable. The calibration results can differ by 20% between two calibration procedures. The maximum deviation recognised during this study was 74%, which is not acceptable. Accelerometers, for example, show a deviation below 1% within one year.
Additional problems recognised within this study are the considerable amount of channels, the effort for the calibration and the durability of the wiring. Comparing these problems with the sensor reliability shows that they are less severe.
In principle, the number of channels could be significantly reduced by electrical combination of single sensors. In the literature, different load limits for the upper middle and lower part of the abdomen as well as for the left and the right side were discussed. Taking these regions into account the number of channels could be reduced to nine (three rows and three columns). The main effort involved in the calibration of the sensors is their removal from the dummy, and the following reapplication of the sensors to the dummy. It is necessary to perform the calibration when the sensors are mounted at the abdomen, in order to reduce the calibration effort. It is then necessary to analyse the quality of this procedure.
The durability issues are limited to the wiring at the abdominal surface. The compression of the parts of the abdominal insert leads to high tension of the cables. To avoid damage caused by the tension, it is possible to pass the wiring in curves. Another option could be the use of more flexible wiring.
In summary, the measurement principle is good, but better sensors need to be developed.

CONCLUSION

In automotive accidents liver, spleen and kidneys are the most affected abdominal organs. All these organs are solid ones. Usually abdominal injuries in frontal accidents are induced by the belt system or the steering wheel. The steering wheel is mainly responsible for injuries of unbelted drivers. Injuries caused by both the lap belt and by the shoulder belt were reported. There are some indications that the

airbag also causes abdominal injuries, but because of the low number of cases analysed in the literature no clear evidence exists. For children, it is the belt system only (either the car belt or the CRS harness).

Based on biomechanical tests, different injury criteria have been proposed. In addition to compression, velocity and force combinations of either of them are in discussion. The most promising injury criteria are the force applied to the abdominal surface and the compression (relative deflection) multiplied by the compression rate (the so-called viscous criterion, VC). The comparison of VC and applied force shows that these two criteria often correlate in a linear way.

Taking into account that abdominal injuries can be caused by steering wheel, lap and shoulder belt, as well as airbags, it is necessary to develop a sensor, which is able to cope with all of these loading cases. Due to high accuracy requirements concerning compression measurement in high speed loading cases with low compression (airbag), the measurement of the surface force offers general advantages. Accuracy of compression measurement is necessary because compression velocity has to be derived from the compression.

The FlexiForce foil sensor was selected as a very promising technology, which can be combined to a sensor matrix. The FlexiForce transducers are flexible pressure sensors with a sensitive area with a diameter of 9.5 mm. They can be used with normal crash equipment and behave sufficiently linearly. In comparison to accelerometers and load cells, they are very cheap. For child dummies of the Q series, namely the Q3 and Q6, 20 of these force sensors cover the entire abdominal insert. In addition to these two dummy sensors, another sensor was developed for the 50th percentile Hybrid III adult dummy using 26 sensors.

Finally, the child dummy sensors were used in a number of detailed accident reconstructions using complete vehicles and fully instrumented dummies within the EC funded CHILD project. The number of cases does not allow a final statistical analysis. However, investigations indicate that the average surface pressure correlates well with injury severity. Additional tests are necessary.

The chosen FlexiForce sensor does not fulfil basic requirements with respect to reliability and repeatability. A considerable change in sensor sensitivity was recognised within a short period of time. Therefore, it is necessary either to improve the sensors used with respect to their performance or to develop new sensors.

ACKNOWLEDGEMENTS

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