

Article

Injury Criteria for Vehicle Safety Assessment: A Review with a Focus Using Human Body Models

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Abstract: This paper aims at providing an overview of the most used injury criteria (IC) and injury metrics for the study of the passive safety of vehicles. In particular, the work is focused on the injury criteria that can be adopted when finite element simulations and Human Body Models (HBMs) are used. The HBMs will result in a fundamental instrument for studying the occupant's safety in Autonomous Vehicles (AVs) since they allow the analysis of a larger variety of configurations compared to the limitations related to the traditional experimental dummies. In this work, the most relevant IC are reported and classified based on the body segments. In particular, the head, the torso, the spine, the internal organs, and the lower limbs are here considered. The applicability of the injury metrics to the analyses carried out with the HBMs is also discussed. The paper offers a global overview of the injury assessment useful to choose the injury criteria for the study of vehicle passive safety. To this aim, tables of the presented criteria are also reported to provide the available metrics for the considered body damage.

Keywords: injury criteria; injury metrics; safety assessment; passive safety; human body models



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1. Introduction

Fatalities related to motor vehicle crashes continue to be, even in the 21st century, a relevant public health issue in all countries. Injuries related to road traffic are considered to be the eighth leading cause of death globally [1]. Furthermore, a report by the Centers for Disease Control and Prevention [2] stated that motor vehicle crashes are the second leading cause of all hospitalizations related to Traumatic Brain Injury (TBI), and result in 18.7% of TBI-related deaths. However, not only the head, but the whole human body is involved in vehicle impacts and must also be considered. Chest injuries are the principal cause of death in approximately 30% of vehicle crashes [3] whereas internal organ injuries tend to be more critical with respect to bone fracture [4], considering the injury at the time, and its long-term consequences. The most common injuries in the case of frontal impacts concern the lower extremities [5]. This overview clearly demonstrates that the study of vehicle safety and the analysis of occupant injuries need to be continuously addressed.

Over the past decades, experimental impact tests using human surrogates (e.g., human volunteers or post-mortem human subjects, PMHS) and dummies (called Anthropomorphic Test Dummies, ATDs) have been performed in order to understand and predict an occupant's response when involved in traffic accidents [6]. The threat to the life of the occupant is based on injury severity. This has been defined by medical staff who have suggested the use of a trauma injury scale. The most used in the automotive field is the Abbreviated Injury Scale (AIS) [7]. This scale allows the ranking and description of injuries by their severity [8].

The injuries can be evaluated, from an engineering point of view, with injury criteria (IC). It is possible to distinguish two IC categories. The first one is mainly based on indirect metrics of kinematics and physical indicators [8]. These injury criteria are mainly applied

when the ATDs are used, thanks to the sensors (e.g., displacements, accelerations, and forces) with which the dummies are equipped. Considering the ATDs, they were developed to study a specific impact configuration. For this reason, different types of dummies are used to study occupant safety. For instance, the Hybrid III and, more recently, the THOR dummy have been used for frontal impact tests [9] whereas, for side impacts, the EuroSID dummy has been replaced today by the WorldSID dummy [10]. The use of ATDs is therefore subject to limitations that are among the reasons behind the development of Human Body Models (HBMs). The HBMs are Finite Elements (FE) models representing real human anatomy. They have been developed both by researchers and companies. Considering the use of HBM and the type of information that can be collected by carrying out experiments on PMHS, which are not only kinematics based, the second IC category is based on strain indicators (plastic strains, principal strain, stresses, pressure distribution) [8]. Therefore, in the automotive field, the HBMs can be used for detailed investigations of occupant injuries.

In addition, with the advent of more and more Autonomous (or Automated) Vehicles (AVs), in which the attention of the driver can be moved away from the road, new and unconventional body postures can be adopted by the occupants [11]. Depending on the automation level of the vehicle, the driver and the other occupants of the vehicle assume a very wide range of postures (i.e., lying down, face to face with the other occupants, with the back or one side of the body towards the driving direction). It is necessary, therefore, to understand the effect of those postures on occupant safety, interaction with the standard restraint systems, and the need for new solutions. From this perspective, whereas the ATDs are aimed at studying the injuries in standard seated positions, the HBMs can be moved and positioned like the real human body. Moreover, the HBMs can be scaled and adjusted to replicate occupants of different sizes, ages, and weights. Therefore, HBMs are very suitable to study occupant safety in AVs. The use of HBMs also implies a change in the evaluation of the injuries with respect to those based on traditional ATDs: strain-based indicators can be used.

In this scenario, the goal of this work is to provide an overview and a summary of the injury metrics that can be adopted for the evaluation of occupant safety, with particular reference to the use of the HBMs. For each body segment, the most known and adopted IC, applicable to the HBMs, are examined and their threshold values are defined. The work is based on a wide review of the literature where scientific papers, books, proceedings of the most important international conferences, and technical reports dealing with the evaluation of injuries, mainly in vehicle applications, were considered and carefully analyzed. The literature review allowed us to define which are the most used IC in the numerical investigations with HBMs compared to the experimental analysis carried out with ATDs. Therefore, both studies carried out on PMHS, and model-based are considered. The proposed limit values can be used to estimate injury probability. In the following, each section of the paper deals with one body region. The segments of the human body are examined from the head to the lower limbs. For each body segment, an overview of the most known IC adopted, the threshold values for the biomechanical parameters, and a short discussion on the most useful metrics for the evaluation of the injuries with particular reference to the use of HBMs are proposed. The work tries to highlight the benefits of the use of HBMs for injury evaluation, discussing the additional information that can be obtained using the HBMs with respect to the results obtained in the experimental tests with the traditional ATDs.

2. Head

Head injuries are one of the most frequent and severe injuries sustained by road users in traffic accidents and account for approximately 40% of road fatalities in the European Union (EU) [12].

In this section, the most relevant injury metrics are presented, at first considering the head as a whole, and then entering into the details of its main parts: the skull and the brain.

Nowadays, the most widely adopted method to assess head injury risk in road safety analysis is the Head Injury Criterion (HIC). It was introduced in order to study head injuries in crash testing by the National Highway Traffic Safety Administration (NHTSA) in 1972, based on the pioneering studies developed at the Wayne State University [13]. This criterion was developed based on previous studies. Gurdjian et al. [14–17] believed linear acceleration to be the most relevant cause of injury and Ommaya et al. [18] supposed that the head injuries produced by direct impacts and linear accelerations were much higher with respect to those produced by head rotation. The HIC is evaluated as the integral of the resultant linear acceleration of the head center of gravity. The integral is weighted to consider both the amplitude and the duration of the sustained accelerations. However, as many researchers pointed out [19–21], the HIC only considers the linear accelerations of the head. Holbourn [22] was the first to state that head angular acceleration was an important cause of injury. Moreover, he assumed that cerebral concussion could be caused by shear and tensile strains produced into the brain by head rotation. Ueno and Malvin [23], and DiMasi et al. [24] stated that the examination of only linear or rotational accelerations may underestimate injury severity. This assumption was also confirmed by Zhang et al. [25], who purported that both the linear and angular accelerations have to be considered to estimate the occurrence of brain injuries. Gennarelli [26–28] confirmed that concussive injuries, such as Diffuse Axonal Injuries (DAI) and Subdural Hematomas (SDH), are more probable to be caused by rotational acceleration rather than linear acceleration. DAI generally occur when high head acceleration and deceleration cause “shear” damage to brain axons [29], while SDH represents a type of intracranial hemorrhage [30]. King et al. [31] also declared that angular acceleration is more damaging to the brain than linear. Willinger et al. [32] confirmed that the HIC, introduced in 1972, has a poor correlation with real-world observations because it does not consider rotational accelerations and impact direction, since it was initially only created for frontal impacts.

With the introduction of FE models representing real human body anatomy, the stress and strain metrics can be used to analyze occupant responses during vehicle impact. Some examples of FE head models have been developed over the years [21,31,33–43].

Consequently, different studies were performed with the aim of establishing new IC able to predict, in a more realistic way, the skull and brain injury risk. From this perspective, Willinger et al. [32] related primary head damages with measurable mechanical parameters: skull fracture can be predicted by analyzing skull deformation; the Subdural and Subarachnoidal Hematoma (SDH and SAH, respectively) are linked to the relative motion between brain and skull; the DAI is mainly related to intracerebral strains and stresses.

Takhounts et al. [44] used a FE model of the human head, to establish physical injury metrics for various types of brain damage, based on stress and strain analysis. Their model was validated against resulting data and statistical parameters obtained with both experimental campaigns and numerical simulations (brain response datasets). Furthermore, they developed the Kinematic Rotational Brain Injury Criterion (BrIC), which, together with the previously introduced HIC, is able to capture most brain injuries and skull fractures. The BrIC is defined as the sum of the maximum angular velocity and the maximum angular acceleration, respectively, divided by corresponding critical values. The critical values are defined by [44]. The BrIC formulation was updated based on the analysis of two different FE head models: the Simulated Injury Monitor (SIMon) [45] and the Global Human Body Models Consortium (GHBMC) v. 3.5 50th percentile [46]. With respect to the original formulation, the updated BrIC also considers different critical values of maximum angular velocity and introduces different components along the main axes of the head reference system. In particular, the updated BrIC is defined as the square root of the sum of the components of the angular velocities divided by their critical values. The new BrIC shows a higher correlation with the occurrence of DAI [47], which is considered to be the most common pathology of TBI, and it is characterized by the dynamic tensile elongation of axonal fibers and consequential rupture [48].

Miyazaki et al. [49] developed a criterion for Rotational Brain Injury (RBI), considering the effects of both the rotational direction and duration of head acceleration. The FE head of the Toyota Total Human Model for Safety (THUMS) ver. 4.0.2 AM50 [50] was used for this purpose.

Between the injury criterion related to human-computer models, more recently Bastien et al. [8] proposed an energy base IC named Peak Virtual Power (PVP). This criterion is derived from the rate-dependent form of thermodynamics, assuming that the injury is represented by the irreversible work in the human body. The PVP is proportional to the maximum rate of entropy during the collision and represents the trauma severity. It is evaluated by multiplying the stress and the strain rate in each element of the organ and the maximum value is taken. The PVP allows predicting the location of the trauma which cannot be evaluated using the strain-based method on human computer models and it can be related to the AIS. It has been demonstrated that the PVP can predict the severity of the injury with high reliability (90%) compared to the AIS, for the injuries of all body regions (brain, skull, thorax, spine, upper and lower extremities).

2.1. Skull

The main damage to be considered when dealing with skull injuries is the cortical bone fracture.

Deck and Willinger [51] showed that the occurrence of skull cortical bone fractures (50% risk) can be related to the strain energy threshold of 865 mJ. Ten years later, Willinger et al. [32] studied a new injury metric based on the Strasbourg University Finite Element Head Model (SUFHEM) [38] and updated the strain energy threshold to 439 mJ. Mattos et al. [47], instead, used the Maximum Principal Strain (MPS) as a metric to assess the presence of a head injury: the MPS maximum value supported in the skull cortical bone was shown to be 0.6%; after that, a skull injury occurs.

2.2. Brain

King et al. [31] discovered that head injuries typically result from direct impact to the head itself or from indirect impacts, such as accelerations applied to the neck and then, by inertia, to the head. In the second case, if the head sustains both linear and angular acceleration, there may be no skull fractures, but still brain injuries.

Ward et al. [52] estimated different threshold values for the Intracranial Pressure (ICP) to assess brain contusion: an injury was detected for ICP values greater than 235 kPa, whereas minor or no injury was assumed to occur for ICP values lower than 173 kPa. The studies carried out by Newman et al. [53] showed, instead, an ICP limit value of 300 kPa for the occurrence of the TBI.

The shear stress metric applied to the whole brain mass can also be used to assess brain injuries. Kang et al. [38] estimated severe brain injury occurrence for shear stress values included between 11 and 16.5 kPa, while Anderson et al. [54] identified severe brain injury (mild DAI) for shear stress values included between 8 and 16 kPa.

Bain and Meaney [55] defined a MPS limit value to avoid damage to the brain white matter: the best overall threshold was defined by the strain value of 21%.

Kimpara and Iwamoto [56] proposed two different injury criteria based on head angular acceleration to evaluate TBI occurrence. In particular, the Rotational Injury Criteria (RIC) was introduced to predict TBI. The RIC is computed as the integral of the angular acceleration measured in the center of gravity of the head. As it happens for the HIC, the integral is weighted to consider both the amplitude and the duration of the sustained accelerations. The RIC results are significantly correlated with the Cumulative Strain Damage Measure (CSDM) with strain thresholds of less than 15%. The CSDM is a useful parameter to predict the occurrence of head injuries using FE head models. It indicates the volume percentage of head elements that ever exceed a given threshold (0.1–0.3 principal strain) during the impact [47]. The second injury criteria introduced by Kimpara and Iwamoto [56] is the Power Rotational Head Injury Criterion (PRHIC), which is strongly

related to the CSDM with strain thresholds greater than 20%, and it is used for more severe TBI prediction. The PRHIC is measured as the maximum value of the weighted integral of the angular component of Head Injury Power (HIP). HIP is defined as the sum of the head mass multiplied by the linear acceleration and the head moment of inertia multiplied by the angular acceleration, as reported by [56].

Gennarelli [57,58] classified DAI severity on the basis of coma conditions and duration: in mild DAI, the coma lasts 6–24 h, in moderate DAI, the coma lasts longer than 24 h, but without abnormal posturing, while in severe cases of DAI, the coma lasts longer than 24 h with brainstem impairment [59]. Deck and Willinger [51] proposed some tolerance limits for the 50% risk of mild and severe DAI injuries. In particular, the brain Von Mises strain threshold was considered to be 25% for mild DAI and 35% for severe DAI. The brain First Principal Strain (FPS) limit value was assumed to be 31% for mild DAI and 40% for severe DAI. The brain Von Mises stress tolerance limit was shown to be 26 kPa and 33 kPa for mild and severe DAI, respectively. The DAI 50% risk threshold, equivalent to the Abbreviated Injury Scale—AIS [60] 2+ injury, was then updated to 27 kPa, based on the SUFHEM head model [32].

Takhounts et al. [41] showed that the 50% probability of DAI corresponds to a MPS value of 87%.

Davidsson et al. [61] suggested, based on animal experiments, thresholds for DAI occurrence in humans exposed to sagittal plane rearward rotation. The proposed limit values were an acceleration of 1000 rad/s^2 with a duration of 4 ms or an angular velocity change of 19 rad/s .

The Minimum of Cerebrospinal Fluid (CSF) Pressure (MCSFP) can be used as a metric to predict the 50% risk of occurrence of Subdural Hematomas (SDH). The limit value was set to -135 kPa [51]. A negative brain cavity pressure can be due to CSF leakages from the spinal canal and lead to intracranial hypotension. The same threshold was defined by Willinger et al. [32] for the CSF internal energy, based on the analyses performed on the SUFHEM head model.

2.3. Discussion

The kinematic-based IC are traditionally adopted when the ATDs are used. In these dummies, the head is treated as a rigid body, and it is therefore easy to measure the physical parameters such as velocities and accelerations. Derived values are then compared with limit values, assuming their relationship with real injuries. The use of the HBMs allows studying of the injuries directly in the area of interest of the head, applying the criteria of medical traumatology and the strain base IC. As a consequence, an evaluation of the head injuries with the HBMs could be made by comparing the measured parameters to the threshold values, as discussed in the previous sections. Opposite to the ATDs, the measure of parameters like acceleration and force, which are still of interest and take into account a half-century of experience, can result in more difficulty than using the HBMs, since the head is not a rigid part as discussed, for example, by Arosio [62]. Therefore, it is necessary to define an equivalent modeling technique for the measurement of physical entities. Normally, the kinematic-based IC do not consider this aspect. Last but not least, there is not a unique worldwide IC recognized as a reliable reference to evaluate head injuries. Moreover, the newly developed IC, that are based on traumatology, can ask for an analysis of mechanical quantities in the form of field distribution and not only as punctual synthetic values.

The injury metrics for the head body segment, discussed above, are summarized in Table 1.

Table 1. Summary of the main IC proposed for the head and the brain.

	Damage	Metric	Threshold	Reference
Skull	Cortical bone fracture	Maximum Principal Strain	0.6%	Mattos et al. [47]
	Cortical bone fracture (50% risk)	Strain Energy	865 mJ	Deck and Willinger [51]
		Strain Energy (SUFHEM-based IC)	439 mJ	Willinger et al. [32]
		Contusion	Intracranial Pressure (ICP)	>235 kPa
Brain	Mild Traumatic Brain Injury (mTBI)	Intracranial Pressure (ICP)	>300 kPa	Newman et al. [53]
		(RIC) Cumulative Strain Damage Measurement (CSDM)	<15%	Kimpapa and Iwamoto [56]
	More severe Traumatic Brain Injury (TBI)	(PRHIC) Cumulative Strain Damage Measurement (CSDM)	>20%	
	Diffuse Axonal Injury (DAI) 50% risk	Von Mises Strain	25% (mild) to 35% (severe)	Deck and Willinger [51]
		First Principal Strain	31% (mild) to 40% (severe)	
		Von Mises Stress	26 kPa (mild) to 33 kPa (severe)	Takhounts et al. [41]
		Maximum Principal Strain (MPS)	87%	
		Von Mises Stress (SUFHEM-based IC)	27 kPa	
	Diffuse Axonal Injury (DAI)	Angular acceleration—duration time	10,000 rad/s ² 4 ms	Davidsson et al. [61]
		Angular velocity change	19 rad/s	
	Brain White Matter contusion	Maximum Principal Strain (MPS)	21%	Bain and Meaney [55]
	Subdural Hematomas (SDH) (50% risk)	Minimum of Cerebrospinal Fluid Pressure (MCSFP)	−135 kPa	Deck and Willinger [51]
Cerebrospinal Fluid (CSF) Internal Energy (SUFHEM-based IC)		−135 kPa	Willinger et al. [32]	

3. Upper Body and Ribcage

In the case of vehicle crashes, the most relevant injuries and fractures of the occupant's upper body are related to the sternum and ribs. The thoracic deflection due to anterior chest loading is the best physical correlation with rib and sternal fractures, in particular, in frontal impacts [63]. Here in the following, a list of the main analyses and studies related to ribcage injuries is reported.

Chest injuries were studied by Forman et al. [64] by means of FE HBMs. They developed a probabilistic method to predict the occurrence of rib fractures. It was assumed that the ribcage injury occurs when the local strain values of the rib cortical bone overcome the ultimate strain. In the same study, fracture initiation was supposed to occur when every single element exceeds the given strain threshold. Consequently, the complete fracture was defined as a structural failure of several adjacent elements.

Bostrom et al. [65] evaluated the risk of rib fractures by considering the chest deflection and rib strains. Chest deflections were measured in 11 locations: at the mid sternum and at the left and right sides of the 3rd–7th ribs.

Hayes et al. [66] evaluated chest deformation by means of chest bands. The bands were inserted in the upper, middle, and lower chest, at the level of the 4th, 6th, and 8th rib. The aim of the analysis was to employ the upper and middle chest bands in the injury assessment of frontal impacts, and the middle and lower chest bands in the analysis of lateral impacts.

Kitagawa and Yasuki [67] studied the correlation between seatbelt loading and mid-sternum deflection to estimate the occurrence of chest injury. The THUMS ver. 4 family (AF05, AM50, and AM95) was used for this purpose. The deflections of the mid-sternum and the rib fractures were analyzed, together with the principal strain of the internal organs.

Golman et al. [68] used the element deletion method to detect rib fractures using the THUMS. The cortical bone plastic failure strain was set to 0.89%, whereas the trabecular bone ultimate failure strain value was set to 13%.

Poulard et al. [69] developed a method to predict the main injuries to the ribs, sternum, and clavicles by analyzing the Ultimate Plastic Strain (UPS) of those parts. The proposed threshold for cortical bone fracture varied, according to the occupant's age, in the range between 3% (20 years old) and 0.8% (75 years old).

Miller et al. [70] virtually instrumented the THUMS ver. 4.01 AM50 to study the injury risks. The 50% risk of thoracic injury was based on the Viscous Criterion (VC) (AIS 4+), defined as the maximum product of the velocity of deformation and the relative displacement or compression: the threshold was around 1.7.

Xiao et al. [71] investigated the effect of two different types of seatbelt loads on the chest injury mechanism, by using the GHBM ver. 4.2. The stress and strain were analyzed to determine the risk of damage to the ribcage. The rib fracture risk was classified into four different groups, according to the First Principal Strain (FPS) values: lower than 25% (low injury risk), 25–50% (middle injury risk), 50–75% (serious injury risk), and higher than 75% (high injury risk).

Han et al. [72] evaluated the fracture of the ribs based on the stress analysis of the model. The study was focused on the evaluation of the chest and abdomen soft tissue injury risk in the case of a three-year-old (3YO) child occupant in a child restraint system (CRS). The THUMS (3YO scaled) was used for this goal and the behavior was then compared to the Q3 dummy FE model. The Von Mises Stress was considered a metric for injury prediction. However, it was not easy to predict rib fracture injuries, since the thorax was soft and the ribs were not easy to fracture.

Kemper et al. [73] performed a three-point bending test on the ribs (from 4th to 7th) of an adult and obtained 130 MPa as the limit value before fracture.

Discussion

Chest injuries were evaluated, in past years, mainly by investigating and analyzing fractures of the upper body bones, chest deformation, and the viscous criteria. The fractures are generally recognized by means of the strains of the bone elements, while the chest deflection and the thoracic deformation are evaluated, respectively, by measuring the distance between the sternum and the vertebrae and as the ribcage shape changes during the impact. The strain measurements can be carefully carried out with HBMs, whereas the thoracic deformation can be investigated both with ATDs and HBMs. The second measurement can be carried out with higher precision using HBMs because they exactly reproduce the chest geometry. Typically, nothing is specifically related to the heart or the lungs, which are located inside the ribcage, even if they are essential organs for human life. Usually, considering the kinematic-based IC, only a generic value of the linear acceleration of the chest is used to consider the injuries to the internal organs. A more detailed investigation can be performed using HBMs and energy-based IC, like PVP [8].

The main injury metrics used to assess ribcage damage are summarized in Table 2.

Table 2. Summary of the main IC proposed for the thorax.

	Damage	Metric	Threshold	Reference
Thorax	Chest deflection	Change in length (11 locations)	-	Bostrom et al. [65]
	Thoracic deformation	Chest bands (at 4th, 6th, and 8th rib)	-	Hayes et al. [66]
	Contusion	Ultimate Plastic Strain (UPS)	3%–0.8%	Poulard et al. [69]
		Viscous Criterion (VC) max	-	Miller et al. [70]
	Rib fracture	First Principal Strain	0–25% 25–50% 50–75% >75% Injury risk groups	Xiao et al. [71]
		Local strain > UTS (rib cortical bone)	Several adjacent elements > threshold	Foreman et al. [64]
		Von Mises stress (stress limit before fracture)	130 MPa	Kemper et al. [73]
		Cortical bone plastic failure strain	0.89%	Golman et al. [68]
		Trabecular bone ultimate failure strain	13%	

4. Spine

In this section, the main studies performed to predict the injury risk of the backbone and of the single vertebrae are proposed.

Mattos et al. [47] focused their work on the cervical part of the spine. The peaks and the average values of the axial force, moment, and strain in the cortical bone of the vertebrae were evaluated to assess fractures.

Gaewsky et al. [74] applied strain-based injury metrics to predict fracture occurrence by using the THUMS ver. 4.01 AM50. The imposed MPS threshold to avoid the fracture of the lumbar vertebrae was 1.5%.

Ye et al. [5] analyzed the fracture mechanism and the associated parameters affecting the thoracolumbar spine response. To this end, a simplified FE vehicle model equipped with the THUMS ver. 4.01 was used. The loads were measured in each mid-vertebral cross-section of the thoracic and lumbar vertebrae, and the Lumbar Spine Index (LSI) was computed. The LSI can be computed as the sum of compressive axial force and resultant bending moment, respectively, divided by their critical values. The considered critical values were, respectively, 1305 N and 34 Nm. Moreover, an age-adjusted LSI version was derived by scaling the LSI. The LSI was computed for each vertebra from L1 to L5, in order to normalize all the lumbar spine responses. The LSI limit value between vertebral fracture and no fracture was demonstrated to be 0.6. Moreover, the lumbar spine shear forces were also evaluated. The shear thresholds for fracture events proved to be 373 N in the longitudinal direction and 273 N in the lateral direction. These directions were, respectively, defined as parallel to the sagittal and frontal planes. With their work, Ye et al. [5] demonstrated that the occupants experiencing thoracolumbar fracture underwent higher strain in the trabecular bone with respect to the cortical bone, with a strain limit value of about 1.71%.

Discussion

The IC applied using the ATDs to assess the injuries of the vertebrae are mainly related to the forces applied to the bones, thanks to the loadcell measures implemented between the rigid elements of the dummy which simulate the spine. The use of HBMs makes possible the analysis of the damage at bone level. To this aim, virtual load cells and other modeling techniques can be implemented in each part of the spine to measure the strain and the shear forces between the vertebrae. The limit values and the injury metrics for this body segment are summarized in Table 3.

Table 3. Summary of the main proposed IC for the spine and the vertebrae.

	Damage	Metric	Threshold	Reference
Lumbar Vertebrae	Fracture	Maximum Plastic Strain (MPS)	1.5%	Gaewsky et al. [74]
		Lumbar Spine Index (LSI)	0.6	
		Lumbar spine shear force	373 N (x) 273 N (y)	Ye et al. [5]
Thoracolumbar Vertebrae	Trabecular bone fracture	Strain	1.71%	

5. Internal Organs

When dealing with injury prediction in the case of vehicle crashes, great importance has to be given to the internal organs. From this perspective, HBM results are very useful. Even if the damage to the internal organs and the soft tissues occurs at a lower frequency than bone fractures, it is generally ranked higher in terms of severity [4].

To correctly analyze occupant response, it is also important to consider the posture within the vehicle before the crash. To this aim, Beillas et al. [75] carried out a study to understand how the position and the shape of the internal organs change depending on the occupant's posture. Four postures were analyzed for this purpose, they were named supine, standing, seated, and forward flexed. It was demonstrated that the thorax volume was more affected by the assumed posture than by the volume of the abdominal organs.

As discussed in the upper body section (Section 3) the IC measure for the internal organs is based on the simple evaluation of the linear acceleration. Therefore, here in the following, the different analyses that can be developed on the FE HBMs to assess the possible injuries of the internal organs are reported. One section is dedicated to each of the main internal organs.

5.1. Lungs

The main damage to be considered when dealing with lung injuries is Pulmonary Contusion (PC).

Gaewsky et al. [74] evaluated the PC risk by means of the MPS computed in each element of the FE model of the lungs. The THUMS ver. 4.01 model was used for the numerical simulations. The strain threshold was set to 34.3%.

Arun et al. [76] used a whole-body GHBMC to analyze occupant kinematics and injuries. The nominal strain threshold of 15% was used. The gross injury risk to the lungs was also computed by means of the VC. To this aim, virtual chest bands were implemented in the model.

Han et al. [72] evaluated the injury occurring to the internal organs by relying on strain-based methods. The considered strain injury limit for lungs was 35%. However, the reported value refers to tests on children.

5.2. Heart

A contusion is the most studied injury to the heart. Shigeta et al. [4] used the maximum principal strain and the pressure on the organ surface to determine the risk of contusion: a strain limit value of 30% was determined.

Nine years later, Han et al. [72] confirmed this strain limit for damage to the heart. In addition, the Ultimate Tensile Strain (UTS) of the myocardial tissue was analyzed to determine the risk of damage: a maximum value of 63% was proposed.

5.3. Spleen, Kidney, Liver, Stomach, and Intestines

Shigeta et al. [4] used the THUMS ver. 4.0 to predict internal organ injury and, in particular, to assess the contusion of the stomach and the small and large intestines. The considered maximum strain limit was 1.2%. The THUMS used for the studies showed high compression of the small intestine during impact (values of about 90% were achieved). However, it was shown that the injuries to the internal organ only occurred when a rib fracture was present.

Arun et al. [76] defined nominal strain threshold values of 30, 20, 30, and 65% for the contusion risk of the liver, spleen, kidney, and bladder, respectively.

Han et al. [72] established a liver injury metric based on the stresses measured on the organ itself: the maximum compressive stress interval was considered to be 0.127–0.192 MPa, according to data coming from animal experiments.

5.4. Discussion

The model-based IC created by the use of the HBMs are suitable to analyze the response of internal organs during the impact. As a consequence, the injury assessment of the internal organs is generally based on threshold values, mainly obtained by means of PMHS tests. With reference to the works presented in the previous sections, it is possible to state that the main damage to be considered when dealing with internal organ injuries is tissue contusion. The corresponding injury metric is the strain analysis, in the majority of the studies present in the literature. However, these IC are not time-dependent and therefore do not consider the impulse duration. Moreover, the strain-based methods do not allow for predicting the trauma location. To overcome these limitations, the energy-based IS named PVP can be used [8].

Additionally, the use of HBMs allows a detailed analysis of the effects of the safety belt. As is well known, during an impact, if the safety belt is not correctly positioned on the iliac crests, it can apply high compression to the soft tissues of the abdomen, with consequent high loads and compression to the internal organs [77]. Therefore, HBMs can provide a higher level of information to the study of the submarining effect [78,79]. In these situations, the injuries to the internal organs can be evaluated, again as previously discussed, with the strain analysis of the tissues or with the analysis of volumetric parameters of the internal organs. The parameters used to assess the injuries to the internal organs are summarized in Table 4.

Table 4. Summary of the main IC proposed for the internal organs.

Damage		Metric	Threshold	Reference
Lungs	Pulmonary Contusion (PC)	Maximum Principal Strain (MPS)	34.3%	Gaewsky et al. [74]
			35%	Han et al. [72]
		Nominal strain	15%	Arun et al. [76]
Heart	Contusion	Maximum strain	30%	Shigeta et al. [4]
	Damage to myocardial tissue	Ultimate Tensile Strain (UTS)	30%	Han et al. [72]
		Ultimate Tensile Strain (UTS)	63%	
Spleen	Contusion	Nominal strain	30%	Arun et al. [76]
Kidney	Contusion	Nominal strain	30%	
Liver	Contusion	Nominal strain	30%	
		Maximum Compressive Stress (MCS)	0.127–0.192 MPa	Han et al. [72]
Stomach	Contusion	Maximum strain	1.2%	Shigeta et al. [4]
Small and Large Intestine	Contusion	Maximum strain	1.2%	

6. Lower Limbs

In this section, the main studies performed to estimate the injury risk to the lower limbs are presented. With reference to the lower limbs, the knee-thigh-hip complex and the knee ligaments, the tibia and fibula, and the ankle and calcaneus complex are the most relevant anatomical parts considered.

6.1. Knee-Thigh-Hip Complex and Knee Ligaments

According to Kuppa et al. [80], the injuries to the knee-thigh-hip complex represent about 55% of lower extremity injuries (AIS 2+, [7]). The limit value for the axial femur force prescribed in the NHTSA Federal Motor Vehicle Safety Standards (FMVSS) 208 for the 50th percentile male is 10 kN. The same value was proposed by Morgan et al. [81] and corresponded to 35% of AIS 2+ injuries.

Other researchers (e.g., [82]) proposed different axial femur force thresholds. In addition, they considered not only the maximum value of the force but also its duration in time. Where the knee ligaments are concerned, Viano et al. [83] proposed the occurrence of partial ligament tears at 14.4 mm of relative displacement between femur and tibia bones, while complete ligament rupture at 22.5° mm relative translation. In 1989, Mertz [84] proposed a limit value of 15 mm of relative displacement as the limit value for injury. Arnoux et al. [85] assumed the ultimate strain values of 28% and 24% to assess the damage to lateral and cruciate ligaments, respectively. Mo et al. [86] proved that the injury threshold of the knee joint undergoing medial shear loading varies from 11.4 to 17.6 mm, with an average of 14.3 mm.

6.2. Tibia and Fibula

The injury criteria applied to this body segment are mainly related to bone fractures.

Mertz et al. [82] studied the tibiofemoral joint angles (ranging from 0° to 20°) and proposed a proximal tibia axial force limit of 8 kN against fracture.

Gaewsky et al. [74] evaluated the first principal stresses in the cortical tibia and fibula to evaluate the injury risk. The ultimate stress limit value for tibia and fibula fracture was proposed to be 134 MPa.

In the works carried out by Ye et al. [5], the lateral displacement at the knee joint level, the change of the tibia, and the abduction angle were considered. With reference to the vehicle reference frame, defined in the ISO 8855 International Standard [87], the lateral displacement at the knee joint was defined as the distance, in top view, between the knee position before and after the impact. The tibia angle variation was defined as the tibia inclination angle from ankle to knee, in lateral view, before and after the impact. The abduction angle was instead computed as the angle between the thighs at knee level in the top view. Moreover, the axial forces and the bending moments were evaluated in the cross sections of the tibia and femur. The cross sections were defined as perpendicular to the axis of the bone, in the upper and lower part of the tibia, and in the middle of the femur. The Tibia Index (TI) was computed to assess possible tibia fracture. The TI is defined as the sum of the compressive axial force of the tibia and the resultant bending moment, respectively, divided by their critical values. The considered critical values were 12 kN and 240 Nm, respectively. A TI value of 1 corresponds to a 50% probability of AIS 2+ tibia shaft fracture, while a TI value of 2 indicates the 100% probability of fracture occurrence. In the same study, the influence of the legs in Out-of-Position (OOP) on the TI was shown.

6.3. Ankle (Pilon) and Calcaneus Complex

Yoganandan et al. [88] observed fractures to calcaneus and tibia distal for tibia axial forces greater than 6.7 kN (50% probability of injury risk), while Bageman et al. [89] proposed a limit value of 7.59 kN for the average tibia force to avoid ankle and calcaneus fracture. Kitagawa et al. [90] proposed threshold values of average tibia load of 8115 N for calcaneal fractures and 7293 N for pilon fractures. Gaewsky et al. [74] used the stress

metric to establish calcaneus and talus injury criteria: the ultimate stress limit value was fixed at 175 MPa.

6.4. Discussion

The analysis of the injuries occurring to the lower limbs during vehicle impact is generally related to the knee, femur, and tibiae. In particular, knee damage can be evaluated by computing the forces which it undergoes or, considering the ligament's failure, by evaluating the relative displacement between the femur and tibia bones. The bone fractures can instead be evaluated by analyzing the cortical bone stress and strain. The use of HBMs allows a detailed analysis of the injuries of the lower limbs, examining the stresses of the bones, whereas, for knee ligaments, the evaluation of their strain can be computed. With ATDs, only the forces and the relative displacement between lower limb segments can be evaluated. The metrics used to assess lower limb injuries proposed in the previous sections are here summarized in Table 5.

Table 5. Summary of the main IC proposed for the lower limbs.

	Damage	Metric	Threshold	Reference
Knee-Thigh-Hip Complex	Fracture	Maximum force (35% risk–AIS2+)	10 kN	Kuppa et al. [80]
		Axial femur force (50th percentile male)	9070 N	Mertz et al. [82]
Knee Ligaments	Partial tears	Relative displacement femur-tibia	14.4 mm	Viano et al. [83]
	Complete failure		22.6 mm	
	Failure	Medial shear loading	11.4–17.6 mm (average 14.3 mm)	Mo et al. [86]
	Lateral ligament failure	Ultimate strain	28%	Arnoux et al. [85]
	Cruciate ligament failure		24%	
Tibia and Fibula	Fracture	Tibia Index (TI)	<2	Ye et al. [5]
		Ultimate Tensile Stress (UTS)	134 MPa	
Ankle (Pilon)—Calcaneus Complex	Fracture	Average Tibia force	7590 N	Begeman et al. [89]
	Fracture (50% risk)		6.7 kN	Yoganandan et al. [88]
	Pilon fracture		7293 N	Kitagawa et al. [82]
	Calcaneus fracture		8115 N	
	Talus fracture	Ultimate Tensile Stress (UTS)	175 MPa	Gaewsky et al. [74]

7. Conclusions

This paper deals with the main injury metrics developed to assess damage to the human body with particular reference to vehicle safety. The most relevant criteria are classified and discussed on the basis of human body segments. In particular, from the top to the bottom of the human body, the head, the chest, the spine, the internal organs, and the lower limbs are examined. Both the injury criteria based on physical testing with anthropomorphic test dummies and the model-based ones, which need the use of finite element human body models, are considered and discussed. The focus is on the second category, which allows the possibility of studying non-conventional postures that can be assumed by the occupants of vehicles during autonomous driving. Moreover, the human body models can be scaled to different sizes, ages, and weights. The work highlights that

the injury criteria based on the use of finite element human body models are more detailed and precise because they also allow for localizing of the injury.

The paper offers a compact and global overview of injury assessment that can be useful in choosing the best injury criteria for the study of vehicle passive safety. To this aim, the work also summarizes the threshold values used to establish the injury risk for each single body part that is examined. Both kinematic-based IC and stress-strain-based IC were considered. With particular reference to the study of safety in autonomous vehicles, the loading mechanism applied to the occupants could change in the same type of accident due to the occupant's position. Therefore, both kinematic-based and stress-strain-based IC should be considered in injury evaluation depending on the body segment under investigation. The IC threshold values could be influenced by the age and the BMI of the occupants, therefore, the appropriate values have to be defined and considered, where a simple scale factor could not represent the ideal solution.

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