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Pedestrian Shoulder and Spine Kinematics
and their Influence on Head Kinematics

RUTH PAAS

Department of Applied Mechanics
CHALMERS UNIVERSITY OF TECHNOLOGY
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Department of Applied Mechanics
Chalmers University of Technology
SE-412 96 Gothenburg
Sweden
Telephone +46 (0)31 7721000

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AND THEIR INFLUENCE ON HEAD KINEMATICS

RUTH PAAS
Division of Vehicle Safety, Department of Applied Mechanics
Chalmers University of Technology

ABSTRACT

Pedestrian to vehicle crashes still represent a major health issue worldwide. In order to prevent fatalities and injuries during the in-crash phase of a pedestrian accident, a sound understanding of pedestrian in-crash kinematics is required and Human Body Models (HBMs) are valuable tools when studying such kinematics. Since injuries to the head feature frequently in severe and fatal injuries in these types of crashes, it is crucial to study the head boundary conditions that govern the kinematics during the crash.

This thesis is focussed on the influence of pedestrian shoulder and spine kinematics and the effect elbow and shoulder impacts have on head linear and angular kinematics. A new full-scale experiment was carried out to establish 3D linear and 3D angular displacements of the head, spine, and both scapulae. Three past full-scale experiments were re-analysed to establish spine curvature. A comparison between the responses of a finite element pedestrian HBM, the Total Human Model for Safety (THUMS) Version 4, and previous shoulder impact experiments comprising volunteers and PMHSs were made. The results were used to study shoulder impacts displaying boundary conditions similar to shoulder impacts in full-scale pedestrian experiments.

The recent full-scale pedestrian experiment provides novel and valuable 6 degrees of freedom (DOF) kinematics suitable for use in HBM evaluations. Head kinematics are governed by neck flexion which is controlled by torso flexion and head inertia. When the pelvis impacts the vehicle, torso flexion away from the vehicle is introduced, influencing the head kinematics. Elbow impacts have considerable influence on head kinematics and elevate the position of the shoulder before shoulder impact. The head impacts the vehicle shortly after the shoulder. Since the load transfer from shoulder impact to the head takes longer than the time between shoulder and head impact, shoulder impacts does not influence head kinematics before the first contact of the head with the vehicle.

The load transfer from shoulder impacts into the torso is governed by the scapular motion over the thorax. Limited scapular motion increases the overall shoulder stiffness and leads to higher head linear and angular displacements compared to lower shoulder stiffness. Elevated shoulder positions limit scapular motion over the thorax while shoulder impacts from supero-lateral directions, which were observed in the pedestrian full-scale experiments, slightly increase scapular motion over the thorax.

THUMS Version 4 is suitable for studying head linear displacements following shoulder impacts although THUMS compares better to tense volunteers than relaxed volunteers. In THUMS, head rotation around the anteroposterior axis is slightly lower and head twist is considerably higher than seen in the volunteers.

KEYWORDS: pedestrian, kinematics, head, shoulder, spine, pedestrian accidents, PMHS, Human Body Model, THUMS

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Paas R, Davidsson J, Masson C, Sander U, Brolin K, Yang JK (2012) *Pedestrian Shoulder and Spine Kinematics in Full-Scale PMHS Tests for Human Body Model Evaluation*, Proceedings of IRCOBI Conference, Dublin, Ireland, p. 730-750.

Division of work between authors: Paas and Davidsson made the outline of this study. Masson performed the new PMHS test and provided material and facilities for the experiment. Davidsson and Paas designed and assembled additional instrumentation and data recordings. Sander provided the accident data. Paas analysed and presented all data included in the paper. The paper was written by Paas with the help of Davidsson, and was reviewed by all authors.

Paper II

Paas R, Davidsson J, Brolin K (2013) *Evaluation of THUMS head and shoulder response in view of application for pedestrian accident analyses*, submitted to Traffic Inj. Prev.

Division of work between authors: Paas made the outline of this study with support of Brolin and Davidsson. Paas re-analysed data from previous experiments and produced all simulation results. Brolin provided support with simulation questions. Preparation of result presentation was done by Paas. The paper was written by Paas, and reviewed by all authors.

Conference Presentations of the Present Work

Paas R, Davidsson J, Masson C, Sander U, Brolin K, Yang JK *Pedestrian Shoulder and Spine Kinematics in Full-Scale PMHS Tests for Human Body Model Evaluation*, IRCOBI Conference, Dublin, Ireland, 2012

DEFINITIONS AND ACRONYMS

AC joint	Acromioclavicular articulation, anatomical joint between the acromion and the clavicle
Acromion	Bony process of the scapula in the superior, lateral region of the shoulder, highest bony landmark on the shoulder
AIS	Abbreviated Injury Scale
Anterior	On / towards the front side of the human body, ventral
APROSYS	Advanced Protection Systems – a European Integrated Project within the 6th Framework program
Biofidelity	Human likeness
BLE	Bonnet leading edge as defined by EEVC 98
C1 – C7	Cervical vertebrae
CIREN database	Crash Injury Research Engineering Network database
CG	Centre of Gravity
Distal	Away from the centre of the body
DOF	Degrees of freedom
EEVC	European Experimental Vehicles Committee
Euro NCAP	European New Car Assessment Programme
FE	Finite Element
GIDAS database	German In-Depth Accident Study database
GH joint	GlenoHumeral articulation, anatomical joint between the head of the humerus and the glenoid cavity of the scapula, commonly referred to as “shoulder joint”
HBM	Human Body Model
HIC	Head Injury Criterion
Humerus	Upper arm bone attached to shoulder and elbow; “head of the humerus” or “humeral head” referring to the proximal section of this bone
Inferior	On / towards the lower side of the body, towards the feet
Lateral	On / towards the (left or right) side of the body
Mid-sagittal plane	Virtual vertical plane through the (ideal) mid-section of the body from anterior to posterior, dividing the body into a left and a right side
NASS database	National Automotive Sampling System database
NHTSA	National Highway Traffic Safety Administration
PCDS	Pedestrian Crash Data Study

Pectoral girdle	Shoulder girdle, in humans consisting of three bones; the humerus, the scapula and the clavicle, and five articulations
PMHS	Post Mortem Human Subject, human cadaver
Posterior	On / towards the back side of the human body, dorsal
Proximal	Close to the centre of the body
SC joint	SternoClavicular articulation, anatomical joint between the sternum and the clavicle
SD	Standard Deviation
SH joint	SupraHumeral articulation, also known as subacromial joint, physiological ("false") joint between the coracoacromial ligament and the head of the humerus
ST joint	ScapuloThoracic articulation, also known as scapulocostal joint, physiological ("false") joint between the scapula and the rib cage
Superior	On / towards the upper side of the body, towards the head
T1 – T12	Thoracic vertebrae
TBI	Traumatic brain injury
Throwing distance	Horizontal distance of the pedestrian CG between initial position immediately before impact and final position on the ground
THUMS	Total Human Model for Safety, developed by TOYOTA MOTOR CORPORATION in cooperation with Toyota Central R&D Labs., Inc.
VRU	Vulnerable Road Users, here defined as pedestrians, cyclists and motorised two-wheelers
WAD	Wrap Around Distance, length measurement from the ground to the location of head impact on a vehicle, as defined by EEVC 98. WAD is usually used for the head but can also be defined for other body parts.

1 INTRODUCTION TO PEDESTRIAN ACCIDENTS

Pedestrian to vehicle crashes continue to remain a major health issue throughout the world. The latest status report on European road safety, published by the World Health Organization (WHO) in 2009, showed that 28 % of all traffic fatalities in the WHO European region involve pedestrians (WHO 2009). In other regions of the world, pedestrian fatalities represent an even more serious issue. In some countries, pedestrians and cyclists together account for as much as 75 % of traffic fatalities (WHO 2013). In high-income countries, pedestrian fatalities have decreased over the past 40 years (WHO 2004) although recent US data indicate a stagnation of this trend. A National Highway Traffic Safety Administration (NHTSA) report (Chang 2008) points out that while the likelihood of a pedestrian accident has declined, the probability of suffering fatal injuries in such a crash has increased. The reason for this finding may be vary; however, it indicates that additional research is needed to reduce the number of pedestrian accidents and reduce the severity of injuries suffered in the remaining accidents.

In order to prevent fatalities and injuries during the in-crash phase of a pedestrian accident, sound understanding of the complex interaction between the pedestrian and motor vehicle is required. In this chapter, an overview is given of pedestrian accident epidemiology, biomechanics and injury mechanisms. State of the art countermeasures are presented, as well as test tools for evaluation of these and future protection systems with a focus on Human Body Models (HBMs). Finally, further in depth investigations into some of the boundary conditions for pedestrian head kinematics from a human body perspective are outlined in this chapter.

1.1 EPIDEMIOLOGY

Numerous studies have been conducted in the past to statistically investigate real-life pedestrian accident data by analysing accident databases. These statistics are essential for the understanding of pedestrian biomechanics and injury patterns. When attempting to improve pedestrian safety through means of experiments or simulation, statistics have the potential to aid focussing on a realistic setup in terms of vehicle speed and braking behaviour, pedestrian stance, and pedestrian avoidance maneouvers, to name just a few.

The vast majority of pedestrians involved in accidents (71-79 %) are standing upright and moving across the road (Maki et al. 2003). Hardy (2009) reports that 65 % are walking and 20 % are running; walking speeds of 0.9-1.7 m/s are reported (Simms and Wood 2009). The amount of pedestrians struck laterally varies between 65 % (Hardy 2009) and 89 % (Simms and Wood 2009). A study by Jarret and Saul (1998) revealed that 60 % of pedestrians made no avoidance maneouver such as jumping or turning away.

Of the vehicles included in the study by Hardy (2009), 66-82 % were passenger cars. They report that the majority of pedestrians was struck by the vehicle front (60-77 %), with 50 % of all pedestrian accidents occurring at a vehicle speed below 25 km/h (Simms and Wood 2009), 70 % at 40 km/h or less (Otte 1999) and 90 % below 50 km/h (Simms and Wood 2009). It is commonly known and confirmed by many studies that injury severity, in general, increases with vehicle speed (Rosén and Sander 2009). Simms and Wood (2009) and Fildes et al. (2004) further point out that the severity increases rapidly from around 40 km/h. Driver avoidance maneouvers such as braking and steering affect the pedestrian impact location on the vehicle. While steering changes the impact location laterally on the vehicle, braking introduces braking pitch and thus changes the head impact location height on the car. Hardy (2009) reports that 1/3 of all drivers did not brake and 30 % applied the brakes hard, while Jarret and Saul (1998) report no avoidance by drivers in 40 % of their cases, and that 43 % applied the brakes. The latter also report that in almost half of the cases where the pedestrian was hit laterally, the vehicle was in a straight path of motion.

The majority of pedestrian accidents occur in built-up areas (Simms and Wood (2009), Harruff et al. (1998)), around half of them on or near an intersection (Hardy 2009). Most common accident conditions include a dry road surface, daylight and fine weather (Hardy 2009). The number of accidents peaks in winter months, on mornings and in the afternoons (Hardy (2009), Harruff et al. (1998)). Statistics on when and where pedestrian to vehicle accidents occur thus appear to be in line with pedestrian to vehicle exposure with increased risk in hazardous environments such as intersections, black ice, and early dusk.

With regards to to fatalities, all studies dealing with age distribution find elderly (60+ or 65+) at a significantly increased risk (Simms and Wood (2009), Hardy (2009), Peng and Bongard (1999), Jarret and Saul (1998)). Some studies also see children at a significantly increased risk (Simms and Wood (2009), Hardy (2009)), while other studies suggest that this age group is at significantly lower risk of fatal injuries (Peng and Bongard (1999), Jarret and Saul (1998)). Fredriksson et al. (2010) point out an increasing risk for chest injuries with increasing age; Demetriades et al. (2004) report an increase of severe head and chest trauma Abbreviated Injury Scale 3+ (AIS 3+) with increasing age. A number of studies suggest that males appear to be at slightly higher risk of being involved in pedestrian accidents, as well as being fatally injured in such accidents, than females (Starnes (2011), Simms and Wood (2009), Hardy (2009), Peng and Bongard (1999), Harruff et al. (1998)). However, those studies did not reveal if male pedestrians are at higher risk of fatal injury due to exposure to higher severity impacts, or due to higher injury risk in similar impact conditions.

Most studies worldwide on severe and fatal injuries agree that head injuries are the major cause of death (e.g., Lau et al. (1998), Hardy (2009)) and if not fatal, can be expected to cause long term, high degree disabilities. Table 1.1 compares injury statistics from two studies on pedestrian accidents (Minzuno (2005), Lau et al. (1998)). The left column shows that the head and legs are most frequently injured in pedestrian accidents, excluding AIS 1 injuries such as skin abrasions. The middle column shows which body part was affected in the main cause of death. It is important to note that “multiple injuries” are listed as an additional item in this column. Among single body parts, head injuries are most likely to be the main cause of death in fatal cases. In contrast, the legs account for a relatively large portion of all injuries (left column), but only represent a low percentage of reasons for death (middle column). The right column shows how many of the deceased pedestrians had suffered a principally fatal injury in each body region, i.e., an injury that was estimated to have led to death even if it would have been the only injury sustained. Thus, the middle column lists only one injury per subject as the main cause of death, whereas the right column lists every injury that would have led to death independently of the other injuries. While this table clearly shows that head injuries are the major problem in pedestrian fatalities, chest and abdominal injuries must be addressed as well. Comparing the middle and the right column, it appears that most fatal injuries in the chest and abdomen occur when multiple injuries are named as the cause of death, similarly for fatal injuries of the neck and extremities. As an example, chest injuries were determined to be the primary cause of death in 2.1 % of the cases, while nearly 50 % of the fatally injured pedestrians sustained chest injuries that would have been fatal even if other injuries had not occurred. This divergence can only be explained if most of the principally fatal chest injuries in the right column are included in “multiple injuries” in the middle column. Similar effects can be seen in the abdominal region and, to a lesser extent, in the neck, pelvis and extremities.

Table 1.1: *Injury frequency distribution in body regions for different severities*

	% of all AIS 2-6 injuries (Minzuno 2005)	Fatal cases, % of all reasons for death (Lau et al. 1998)	Fatal cases, % of all subjects, principally fatal injuries (Lau et al. 1998)
Head	31.4	42.0	83
Face	4.2	NA(*)	NA(*)
Neck	1.4	2.1	12.5
Chest	10.3	2.1	49.9
Abdomen	5.4	2.6	40.9
Pelvis	6.3		
Legs	32.6	0.8	10.8
Arms	8.2		
Multiple injuries	NA	49.0	-
Other	NA	1.1	1.1

(*) Facial injuries possibly included in head injuries

Several studies investigate impact locations of the head and other body parts on the car. Hardy (2009) mentions that windscreen and windscreen frame together account for around 30 % of all pedestrian injuries, with around 31 % of all injuries being head injuries. The windshield area has been identified to be the most common site of head impact for adults (Fredriksson et al. (2010), Minzuno (2005), Yao et al. (2008)) and the bonnet for children (Yao et al. 2006). Fredriksson et al. (2010) found that almost two thirds of AIS 3+ head injuries were caused by structural parts in the region of the windshield, indicating that not the windshield itself but the stiff parts in the windshield region are the most hazardous to the head.

Historically, accident statistics as shown in this chapter have had major implications for the approach taken to understand and replicate pedestrian accidents. For simplification purposes, what is known as a standard scenario was defined as a pedestrian in an upright position being hit laterally by the mid-part of a passenger vehicle front while walking and without attempting avoidance. Car speeds around 40 km/h are common in experimental setups, taking into account both the number of accidents and the accident severity. As shown in this chapter, each standard scenario variable alone represents the majority of cases. However, the percentage of cases meeting all of these criteria might be much lower. In order to understand real-life pedestrian accidents and ultimately improve pedestrian safety in real life, it is necessary to investigate more than just the standard scenario.

1.2 Biomechanics and Injury Mechanisms

Typical pedestrian in-crash kinematics have been reported in many studies (e.g., Simms and Wood (2009)) In the standard scenario, the first contact between vehicle and pedestrian is between the bumper and the legs, usually in a region at or around the knee, depending on knee height and bumper height with or without braking pitch. In this phase, an initial rotation around the longitudinal axis of the pedestrian is initiated. The direction of this rotation depends on which leg is impacted first, since this determines the position of the lever arm that triggers the rotation. Some milliseconds later, the Bonnet Leading Edge (BLE) impacts either the pelvic region (sedan type car), the upper leg (sports car or tall pedestrian) or the thorax (Sport Utility Vehicle (SUV) type car or child pedestrian). Up to this point, the upper torso and head are usually still upright. As the vehicle continues its path, the torso begins to rotate towards the vehicle due to the fact that the legs and pelvis are partly in contact with the front parts of the vehicle which continues to travel forwards. While the torso rotates towards the vehicle, the arms do not naturally follow this motion since the shoulder joint can move rather freely. Instead, inertia and gravity keep the upper arms vertical, while the vehicle moves forward and the torso falls towards the bonnet. This sequence of events makes the car impact the pedestrian’s elbow, upper arm and/or shoulder in some cases, e.g., if the upper body rotation around the longitudinal axis is slight. Similar to the upper arm, the head does not follow the motion of the torso immediately either. The atlantooccipital joint provides a rather extended range of motion for the head relative to the torso so that, while the torso falls down towards the car, inertia causes the head to lag behind this motion until the neck finally drags the head towards the car, followed by head impact on the vehicle. After the phase from first contact between the pedestrian and the car to head contact (primary impact), there is either a flight phase, a phase of the pedestrian being carried on the bonnet, or a sliding or rolling off phase. Which of these occurs depends largely on vehicle shape, speed, and braking (Hamacher et al. 2012). A steep vehicle front is likely to throw the pedestrian forward, resulting in a flight phase. A sports car travelling at high speed can result in the pedestrian flying over the car and impacting the ground behind the car, resulting in a different kind of flight phase. Sedan type cars or sports cars at low speeds that brake during impact are likely to result in carrying the pedestrian for a short amount of time and consequently sliding or rolling off the vehicle front (Hamacher et al. 2012). Subsequently, the pedestrian hits the ground (secondary impact) and continues to slide on or roll over the ground until reaching the resting position, with the risk of further impacting structures on or near the road (tertiary impact).

Pedestrian kinematics show that specific areas of the body are at particular risk of getting injured during the primary impact in a pedestrian to vehicle crash (Simms and Wood (2009), Hardy (2009), Carter et al. (2008)). Depending on where the bumper impacts the legs, the long bones of the leg, i.e., the tibia and fibula in the lower leg or the femur in the upper leg, are at risk of fracture due to bending and shearing. Likewise, the knee is at risk of ligament rupture and patella fracture while fractures might occur to the pelvis during impact with the bonnet leading edge and thoracic trauma may be caused by blunt impact with the bonnet for adults or with the bonnet leading edge for children (Carter et al. 2008). Finally, when the head hits the vehicle, there is risk of skull fracture and brain injuries. Possible additional risk of injury to the vertebral column and surrounding ligaments, as well as brain injury might be present in the phase between pelvis and head impact due to the head lag and catching up motion. This type of motion causes head rotation and thus the risk of relative motion between the skull and the brain, which is known to cause traumatic brain injury (TBI). In addition, there is risk for abdominal, thoracic and upper limb injuries due to blunt trauma (Hardy (2009), Carter et al. (2008)).

1.3 COUNTERMEASURES

Early attempts to protect pedestrians were developed shortly after the first passenger vehicles were seen on the roads, mainly focussing on separating pedestrians from passenger vehicles (Fruin 1973). However, modern pedestrian safety efforts improving the vehicle front were not attempted until the 1980s (Simms and Wood 2009) after a number of studies throughout the 1970s had discovered the potential for improvements despite the inequality in mass, speed and stiffness of the collision partners (e.g., Kramer et al. (1973), Ashton (1975), Pritz et al. (1975), Krieger et al. (1976), Bourret et al. (1979)).

More recently, both passive and active safety measures have been developed and combined to form integrated safety for pedestrians. In the area of passive pedestrian safety, one of the earliest improvements was the removal of sharp objects from the radiator grill, the front lights, and the bonnet areas; later, the BLE was rounded off and softened, and softer bumpers were installed (Longhitano 2009). The inclusion of pedestrian safety in the European New Car Assessment Programme (Euro NCAP) rating was another milestone fuelling development of further countermeasures (Simms and Wood 2009). It was especially when Euro NCAP introduced a new and more inclusive rating scheme in 2009 and pedestrian safety was included in an overall safety benchmark of vehicles that manufacturers were awarded for active safety measures (Euro NCAP 2013b). Since then, considerable effort has been made by car manufacturers to improve all aspects of pedestrian safety from a vehicle design perspective (Lindman et al. 2010).

To date, state of the art pedestrian passive safety available on the market includes reduced stiffness of the bumper, BLE, bonnet, A-pillars and the upper windshield frame. Increasing the space for deformation between bonnet and the components below such as the battery has increased pedestrian safety. Increased space for deformation and reduced stiffness have been accomplished, e.g., introducing active bonnets that are raised automatically in case of a pedestrian impact, as well as air bags on the outside of the car, especially in the bonnet rear end and A-pillar regions (Jakobsson et al. 2013). In addition, the vehicle front geometry has the potential to contribute to pedestrian safety. Reducing the space between the lower vehicle front and the ground by lowering the front end or by adding structures such as a secondary bumper or an external airbag prevents the feet from being dragged underneath the car and reduces lateral knee loading (Thollon et al. (2007), Pipkorn et al. (2007)), while lowering and smoothing the BLE curvature reduces pelvic injuries by allowing the pelvis to slide over it (Kallieris and Schmidt 1988). Advances in active safety have enabled vehicles to detect pedestrians and engage automatic braking in case of emergency which has the potential to help avoiding pedestrian accidents or at least mitigate the consequences (Lindman et al. 2010). However, active safety is not expected to be able to prevent all pedestrian accidents in the near future due to mechanical limitations in braking capabilities, limitations in pedestrian detection by sensors, and limitations in predicting pedestrian behaviour. Reducing pedestrian fatalities and injuries by means of passive safety is still necessary and will be an important contribution to pedestrian safety in the future. Although statistics indicate that previous pedestrian protection has already proved significant benefits, i.e., the annual pedestrian fatalities in Europe have been reduced from about 13 000 to about 6 000 between 1980 and 2000 (Breen 2002), pedestrian safety is still a major health issue (WHO 2013).

1.4 Accident Data and Physical Test Tools

Accident data, as well as physical and numerical test tools provide data sources for pedestrian in-crash kinematics. Accident data enabling assessment of pedestrian kinematics require in-depth reconstruction of an accident. Physical test tools are required to be human-like, repeatable, and durable. This chapter provides an overview on accident data sources and physical test tools; an introduction to numerical test tools will follow in Chapter 1.5.

Accident databases such as the German In-Depth Accident Study (GIDAS, Otte et al. (2003)), the Pedestrian Crash Data Study (PCDS, Jarrett and Saul (1998)), and the APROSYS (Advanced Protection Systems) database (Carter et al. 2008) provide statistics on pedestrians' stature, age, behaviour immediately prior to the crash, and injury outcome, as well as information on the vehicle and environmental data. However, matching indentations and damage on the car with specific body parts can be challenging and some variables such as the exact pre-impact posture of the pedestrian can be difficult to reconstruct. Thus, it is often difficult to reconstruct the exact pedestrian in-crash kinematics from real-life accidents, although accident databases provide the only real-life data source for pedestrian accidents, and are thus essential for preventing future fatalities and injuries.

Pedestrian crash test dummies have been in continuous development for several decades. Honda developed the Polar-II and the updated version, Polar-III (Akiyama et al. (1999), Akiyama et al. (2001)), based on the frontal crash test dummy Thor (White et al. 1996). In addition, a pedestrian version of the Hybrid II and the updated version Hybrid III (Humanetics Innovative Solutions 2013) have been developed. Together with HBMs (Chapter 1.5) and PMHSs (Post Mortem Human Subjects, Chapter 1.6.1), crash test dummies are the only available tools to date for evaluating passive safety systems in full body testing before they enter the market. However, designing a robust dummy strong enough to tolerate severe impacts, without breaking, may be a conflicting goal to developing a biofidelic dummy. Pedestrian trajectories in PMHS experiments generally display a considerable amount of spread (Ashton et al. 1983) which presents a problem in the evaluation process of passive safety systems. Therefore, pedestrian crash test dummies facilitate repeatable trajectories although pedestrian dummies that are too simplified will most likely not predict biofidelic trajectories or realistic head impact conditions. This can be seen in non-biofidelic kinematic results of early simplified pedestrian dummies (Simms and Wood 2009). Later versions of the pedestrian crash test dummies, such as the Polar II, are more comparable with the PMHS experiments (Kerrigan et al. 2005b) while the head trajectory remains to be addressed further (see also Chapter 1.6). Nevertheless, physical full-scale pedestrian testing with standardized subjects against a real vehicle is still considered a valuable complement to numerical simulations.

Subsystem impactors have been developed to simplify testing and increase repeatability of test results. Current pedestrian Euro NCAP safety assessments include using adult and child headform, upper legform and legform impactors to assess car front ends (Euro NCAP 2013a). While an obvious goal is repeatable results, especially in regulatory testing, a number of shortcomings have been identified with regards to subsystem impactors. Due to pedestrian Euro NCAP tests being based on relatively simple boundary conditions, the adult head impactor, for example, impacts several points on the vehicle at one impact speed (11.1 m/s) and one angle (65 ° to ground level) only. These boundary conditions do not cover a wide range of potential head impact conditions, whereas head impact speed and angle vary in PMHS experiments (Kerrigan et al. 2008) and simulations (Elliott et al. 2012). The sensors used in the head impactors measure 3D force/acceleration in the head impactor centre of gravity (Euro NCAP 2013a). To date, there is no assessment of head rotation, neck load, or spine curvature, which all pose potential hazards for pedestrian injury (Chapter 1.2). Thus, while regulative testing provides necessary motivation towards improved pedestrian safety, the usage of impactors might not address some of the safety issues present in real-life accidents.

1.5 HUMAN BODY MODELS

In recent years, pedestrian HBMs have been further developed and are continuously being improved. One major advantage of HBMs in general is that they are relatively easy to adjust or modify. Thus, they can be used to study many different impact conditions without causing physical damage to a real-life vehicle, crash test dummy or PMHS which makes them suitable tools during the early stages of product development by car manufacturers, variation studies, and preparation of physical tests.

Until recently, two main approaches to human body modelling have been available for use in the in-crash phase; one is MultiBody (MB) modelling and the other is the Finite Element (FE) method. As both have different advantages and shortcomings (Wismans et al. 2005), combined MB/FE modelling has recently been developed as a third option. MB models are usually composed of rigid body ellipsoids and planes with a point mass in their centres of gravity. These bodies are connected by joints with a lumped parameter joint stiffness, simulating the interaction between bones, muscles and ligaments. Contact and penetration characteristics are approximated by idealised functions. This approach allows for low computation time, while the level of detail in MB models is limited. On the other hand, MB models have been reported to predict overall kinematics with high accuracy, which is valuable for accident reconstruction and the prediction of throwing distances. Due to these advantages, MB models are often a preferred tool in accident reconstructions.

In contrast, FE models consist of deformable elements and can be used to predict injury based on tissue level criteria by calculating variables such as stress, strain, and strain rate. The Finite Element Method (FEM) allows for modelling of complex geometries and using advanced material laws, providing a high level of detail. With FE models, it is possible to study the load path through the human body in an impact on tissue level. Therefore, FE models for pedestrian impacts should be validated both in terms of kinematics and on tissue level.

The Total Human Model for Safety (THUMS) pedestrian model is a commercially available full body FE model developed jointly between Toyota Motor Corporation and Toyota Central R & D Labs., Inc. (Watanabe et al. 2011). Four main releases have been made available to date, version 1, launched in the year 2000 and the latest, version 4, launched in 2010. While a number of studies have attempted to evaluate the biofidelity of THUMS Version 1 (e.g., Maeno and Hasegawa (2001), Iwamoto et al. (2002), Chawla et al. (2005), Pipkorn and Mroz (2009)), a certain amount of full-scale and component level evaluation of THUMS Version 4 was performed by (Shigeta et al. (2009), Watanabe et al. (2011), Watanabe et al. (2012)). Full-scale kinematics evaluation was performed against three PMHS tests, comprising a sedan, an SUV, and a minivan (Watanabe et al. 2012). 2D linear displacements of the head centre of gravity, T1, L5/S1, the knees and the heels were shown to generally match the PMHS results. However, the head impact locations of THUMS were lower than in all PMHS tests, up to around 10 cm for the sedan, as well as the minivan. At component level, Shigeta et al. (2009), Watanabe et al. (2011), Watanabe et al. (2012) evaluated validation steps with regard to impact responses of the head and neck in frontal and lateral impacts, head rotation with respect to brain kinematics and brain injuries, direct impact to the head, chest responses in several frontal and lateral impact conditions, frontal abdominal impact responses, lateral impact and four-point bending responses of the knee, as well as static 3-point bending and dynamic compression responses of the humerus. To the best of the author's knowledge, other validation studies of the THUMS 4.0 pedestrian shoulder and spine were not available in the literature during the period of this study.

1.6 BIOMECHANICAL TESTS

Whilst accident data and crash test dummies (Chapter 1.4), as well as human body models (Chapter 1.5), provide necessary statistics and valuable tools for reconstruction and safety system evaluation, biomechanical tests are crucial for improving the understanding of pedestrian accidents both on component and on full-scale level, as described in this chapter.

1.6.1 FULL-SCALE EXPERIMENTS

Extensive pedestrian full-scale experimental testing began in the 1970s when attention was brought to the increasing number of fatalities and injuries involving pedestrians in accidents with motor vehicles. Kramer et al. (1973) investigated fracture mechanisms of the lower legs in pedestrian impacts, which has been one of the focus areas in pedestrian testing since the beginning. One of the first studies focussing on the whole body was carried out by Pritz et al. (1975) where the influence of vehicle design on pedestrian injury was examined. The study reports the understanding of pedestrian injury mechanisms at that time as follows; pedestrian to car contacts were mainly responsible for a greater amount of severe leg and pelvic injuries, while the pedestrian to ground impact was believed to cause a higher number of severe head and arm injuries. This differs from today's understanding and might partly be due to that the vehicle front shapes were different at that time, and that the study predominantly focussed on leg and pelvic injuries. Investigating a lowered vehicle front end profile, the study highlighted that the overall effectiveness of this countermeasure would depend on its effect on upper body and head injuries. It was found that both lowering and softening the vehicle front end tended to increase head velocity. In addition, they discovered that the head velocity peak was higher than the velocity of the bonnet in all experimental tests and that this peak occurred before head impact during a "whipping, rotation motion about the upper body that suggests a potential for neck injuries" (Pritz et al. 1975). Krieger et al. (1976) studied pedestrian kinematics in detail, focussing on leg and head acceleration but describing body rotations as well. They concluded that pedestrian accidents lead to "a wide variety of complex motions" of the pedestrian. Based on the Head Injury Criterion (HIC), the risk of injury was higher in head to bonnet impact than in head to ground impact, and that "dummy and cadaver response to almost identical impacts were quite different". An early mathematical pedestrian model was introduced by Ashton et al. (1983), a study in which reconstructions of several real-life accidents were attempted using dummy testing, PMHS testing, and a mathematical model. One of their findings was that lowering the vehicle front end reduced head impact velocity, opposite to the findings of Pritz et al. (1975). They too, note the difficulties in reproducing a specific event due to the complexity of the pedestrian kinematics, as well as considerable differences between dummy and PMHS responses. Other notable early full-scale pedestrian PMHS tests have been performed by Cesari et al. (1980), Farisse et al. (1981), Cavallero et al. (1983), Brun-Cassan et al. (1984), Kallieris and Schmidt (1988), and Ishikawa et al. (1993), to name just a few, although they were not used for model comparison in the present study. To reproduce the tests in these studies would have been challenging, partly due to the used vehicles being rather outdated, but foremost since high speed videos and digital data could not be obtained from these tests.

Schroeder et al. (1999) studied kinematics and injuries of six PMHSs that had been impacted by two different vehicles in order to investigate advances in vehicle designs aimed at protecting pedestrians. They concluded that, while a lower vehicle front end was found to reduce leg injuries, head impact to the vehicle remained an issue to be addressed. They also reported spine elongation during the loading of the pedestrian onto the bonnet. They hypothesised that such elongation would have contributed to spine injuries discovered in the subjects following the experiments.

Kerrigan et al. (2005a), Kerrigan et al. (2005b), and Kerrigan et al. (2007) investigated the kinematics of PMHSs and the Polar-II dummy in impacts against two different mid-size sedan cars and an SUV in an attempt to establish kinematic corridors by scaling time, as well as the trajectories for each body segment. One of their findings was that the PMHSs generally showed longer Wrap Around Distances (WADs) than the dummy, which they attribute partly to the PMHSs tendency to slide more over the bonnet than the dummy, and partly to the lack of muscle tension in the PMHSs. The dummy, in contrast, was designed to simulate a greater lateral bending stiffness similar to a living human. While they found the PMHSs' heads lagging behind the upper torso during the upper body rotation over the bonnet towards the vehicle, this effect was not as pronounced in the dummy, which was attributed to the greater neck stiffness in the dummy compared to the PMHSs.

Subit et al. (2008) studied the kinematics and injuries of four PMHSs, where one short and one tall subject was impacted by a small city car respectively and a mid-sized sedan. The study focused on the pelvis and upper body kinematics which were found to depend on the subject size and the vehicle front end geometry. For the taller subjects, they noted an increased amount of sliding over the bonnet. In contrast, the shorter subjects displayed a considerable change in pelvis kinematics around the time of pelvis contact. Subsequently, around the time of head impact, the shorter subjects displayed a higher amount of lateral bending than the taller subjects, and a higher HIC score was measured for the shorter subjects for each of the vehicles. From the post-test injury assessment, it was noted that the subjects impacted by the small city car sustained more rib fractures than the subjects impacted by the mid-sized sedan. It also found all four subjects sustained some form of spinal injury, all of which involved fractures, either to the vertebral body (three subjects) or the processes only (one subject).

The head velocity peak was consistently found to occur before head impact in all examined recent studies which have investigated head velocity curves (Schroeder et al. (1999), Kerrigan et al. (2005a), Kerrigan et al. (2005b)). The head lagging behind the upper torso during rotation over the bonnet was seen in all these studies and with all investigated vehicle front geometries; even though increased neck stiffness appeared to considerably reduce this effect. In order to reduce variability, the hands were attached to each other in most of the studies. Substantial elbow impact was seen in two out of three tests in Schroeder et al. (1999), while in the differing test result, the upper arm was restrained. Shoulder impact was seen in all three of these tests. Both elbow and shoulder impact were seen in the one PMHS test displayed in Kerrigan et al. (2005a), as well as in Kerrigan et al. (2005b) although the hands were tied together in their experiments. In the four experiments carried out for the study of Subit et al. (2008), communication with the authors revealed that all four test subjects impacted the vehicle with the shoulder while only one subject displayed a considerable elbow impact, i.e., elbow impact in which a notable amount of energy can be expected to be transferred into the upper body.

1.6.2 Shoulder Impact Experiments

Shoulder impacts have mostly been studied in relation to side impacts in the past, investigating occupant responses in impacts with the side interior of the vehicle such as the door or the B pillar. The shoulder impact experiments existing to date have, to the best knowledge of the author, all been performed with seated subjects. These experiments can be categorised into, on the one hand, those that use a relatively small surface impactor, most often directed against the glenohumeral (GH) joint (e.g., Bolte et al. (2000), Thollon (2001), Marth (2002), Bolte et al. (2003), Compigne et al. (2004), Ono et al. (2005), and Subit et al. (2010)), and on the other hand those that additionally directly engage other body parts, such as the thorax or the pelvis, as well by using a larger impactor surface (e.g., Cavanaugh et al. (1990), Irwin et al. (1993), Koh et al. (2001)). To study the influence of shoulder impacts on the head kinematics in pedestrian accidents, both these groups of studies are of interest; although when evaluating the shoulder of an HBM, well-defined conditions and impacts locally restricted to the shoulder are preferred for biofidelity assessment.

Bolte et al. (2000) investigated shoulder response and injuries of eleven PMHSs being hit by a 23 kg padded impactor at velocities between 3.5 and 7 m/s in order to establish an injury probability distribution. The study found that tissue thickness of the lateral aspect on the shoulder varied considerably (6.4 mm ± 4.2 mm), that 41 % of the impacts caused a distal clavicle fracture, and that 63 % of the impacts resulted in a loose sternoclavicular joint. Arm support was not provided in these experiments.

Bolte et al. (2003) studied shoulder response and injury performing 14 shoulder impacts to PMHSs with a 23 kg padded impactor at velocities of 4.4 (12 tests) and 7 m/s (2 tests) in lateral, 15 ° anteriorly oblique and 30 ° anteriorly oblique impact directions. One of the main findings was that oblique loading resulted in greater shoulder deflections, which was attributed to the scapula sliding posteriorly over the thorax. In this test series, regarding bone injuries, only one distal clavicle fracture and one subject sustaining four rib fractures were reported, and the most common joint injury, a loose sternoclavicular joint, was found in 38 % of the experiments. The reduced amount of injuries compared to Bolte et al. (2000) was attributed to impacts having been aimed below the injury threshold. In side impact car crashes obtained from the National Automotive Sampling System (NASS) database, they found that the risk for AIS 2+ shoulder injuries is 1.3 %, the majority of which are clavicle fractures. Their search in the Crash Injury Research Engineering Network (CIREN) database revealed that 75 % of all clavicle fractures were fractures of the distal third, caused by B-pillar impact and interior door impact. These numbers indicate that fractures of the shoulder bones are rare in side impact crashes although they point out that the distal clavicle is one of the weakest points in the shoulder bones. Based on three-point bending tests of the clavicle, they determined the average stiffness as 147 N/mm and the average failure load as 680 N.

Compigne et al. (2004) subjected seven PMHSs to lateral and oblique (± 15 °) impacts at 1.5 m/s and purely lateral impacts at 3, 4, and 6 m/s with a rigid 23.4 kg impactor. Similar to Bolte et al. (2003), they found a greater mobility in the shoulder when the impact was lateral or oblique from anterior than when it was oblique from posterior, which, once again, was attributed to the scapula sliding over the rib cage. In addition, for the anterior oblique and the lateral impacts, they found rotation of the clavicle in the horizontal plane around a point close to the sternoclavicular joint assumed to be the result of the impact pushing the scapula backwards over the thorax, and thus pulling on the clavicle. The maximum scapula motion reported in their tests is 5 cm. For the high speed impacts, they found that around 20 % of the impact energy transferred to the subjects was transformed into internal energy, or tissue deformation, while the rest was transformed into kinetic energy, or subject displacement. Fractures occurred in six out of seven subjects; however, the amount of internal energy rose to 30 % in one case where the scapula was fractured. It was reported that fractures of the clavicle or the scapula as well as ligament ruptures lead to an increase of scapula, and thus acromion motion. It is unknown if arm supports were used; the description of the methods used make reference to Compigne et al. (2003) which, in turn, refers to the methods used in Bolte et al. (2000), where no arm support was used.

Other pure shoulder impact studies with PMHSs were performed by Meyer (1994) and Thollon (2001), who impacted PMHSs embalmed in a Winckler’s solution with 37 kg padded impactors at speeds of 5.5 and 7.5 m/s. They found two peaks in the contact force and identified the first one as the moment when the humeral head bumped into the glenoid fossa of the scapula. In the ISO/TR 9790 (1999) document, shoulder impact tests were reported where ±15 ° oblique directions of impact were used and contact force-time corridors were established for impact speeds of 4.5 m/s. Marth (2002) subjected twelve PMHSs to lateral impacts with a 23.4 kg impactor at 4.5 and 6.7 m/s and analysed acceleration, force, and displacements.

Ono et al (2005) subjected eight volunteers to lateral and ±15 ° oblique impacts with an 8.5 kg rigid impactor and pre-defined load curve, calibrated with a Hybrid III dummy to maximum contact forces of 400, 500, and 600 N. The volunteers were asked to relax their muscles in one set of tests and to tense them in a second set. Corridors were established for impact load, head, T1 and pelvis accelerations, neck force and moment, shoulder deflection, and head, T1, and head relative to T1 rotation angles. One of their findings was that the maximum head acceleration did not change considerably when comparing relaxed and tensed volunteers, but that shoulder deflection was reduced by 20 % for the tensed volunteers; the maximum neck moment around the anterior-posterior axis (where the main head rotation occurred) was reduced by 24 %. Differences in head/neck/torso responses were also found due to different impact directions, which they attribute “to the difference in shoulder anatomical shape or structure”. The shoulder motion was found to be restricted especially in the case of posterior impact, comparing well to the findings of the studies named before in this chapter.

1.7 Shoulder and Spine Anatomy and Physiology

The human shoulder consists of three bones, the humerus or upper arm bone, the scapula or shoulder blade, and the clavicle or collarbone, as well as muscles, tendons and ligaments (Marieb and Hoehn 2010), see Figure 1.1. Five articulations or joints, contribute to the motion of the shoulder, three of which are anatomical joints and two of which are false, or physiological, joints.

The major anatomical joint in the shoulder is the glenohumeral joint, a multiaxial synovial ball and socket joint connecting the humerus to the glenoid fossa of the scapula. This joint is commonly referred to as “shoulder joint”. Due to limited interaction between the bony surfaces, it is the most flexible joint in the body and allows for a major part of the upper arm range of motion (Marieb and Hoehn 2010). The acromioclavicular joint is a gliding synovial joint, functioning as a pivot point, forming the articulation between the acromion as part of the scapula and the clavicle which allows for greater arm rotation. The third anatomical joint is the sternoclavicular joint, forming the articulation between clavicle and sternum. This synovial double-plane joint makes movement of the clavicle possible in three planes, enlarging the range of motion of the shoulder even more.

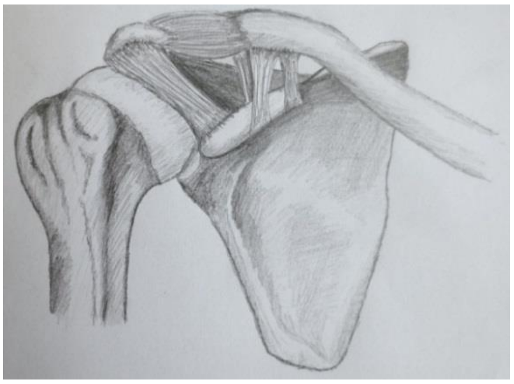

Figure 1.1: The bones of the shoulder, anterior view: humerus (left), clavicle (top/right) and scapula (mid/bottom). Including the glenohumeral capsule, the coracoacromial ligament (left ligament), the acromioclavicular ligament (top ligament), and the coracoclavicular ligament (right ligament in two parts, left: trapezoid, right: conoid)

In addition to these three anatomical joints, two physiological joints are part of the pectoral girdle and thus contribute to shoulder motion. The supra-humeral joint is an articulation of the head of the humerus and the coracoacromial ligament, supporting the glenohumeral joint in providing a greater range of motion. The scapulothoracic joint is the articulation between the anterior scapula and the posterior rib cage, where muscles and tendons allow the scapula to slide over the rib cage and thus allow two translational and one rotational degree of freedom of scapular motion.

These articulations contribute to the shoulder being the human joint with the largest range of motion, while it is also the most complex joint in the human body (Marieb and Hoehn 2010). Measuring the range of motion of single parts of the shoulder has been attempted a number of times in the past (e.g., Pearl et al. (1992), Kebaetse et al. (1999)), but has proven to be difficult since several parts of the shoulder complex contribute to each movement. The overall response of the shoulder to direct impact has been studied in general as well (Chapter 1.6.2); however, the author has not found any evidence that the mechanism of energy transfer through the shoulder into the upper torso and spine has been studied yet.

The human spine normally consists of 24 vertebrae, which form the articulations of the spinal column, and two bones in the lower part, the sacrum and the coccyx (Marieb and Hoehn 2010). The upper seven vertebrae (C1-C7) form the cervical spine. C1 (also called atlas) and C2 (also called axis) contribute to the head range of motion to a major degree. The atlanto-occipital joint allows mainly for a nodding kind of motion, while the atlanto-axial joint allows mainly for rotating the head to the left and right (Marieb and Hoehn 2010). The other cervical vertebrae (C3-C7) allow for flexion, extension, lateral flexion and rotation of the neck. Below the cervical vertebrae there are twelve thoracic vertebrae (T1-T12) which all connect to the ribs. This section of the spine enables rotation in the thoracic region plus a limited amount of lateral flexion, limited by the ribs, and limited flexion and extension. Further below there are five lumbar vertebrae (L1-L5) which have to carry the most weight of all vertebrae and are therefore larger with a more robust structure. The lumbar spine allows for flexion and extension, as well as a limited amount of lateral flexion, but rotation is prevented (Marieb and Hoehn 2010). The sacrum itself does not contribute to the spinal range of motion although limited motion might be possible in the articulation between the sacrum and L5. The sacrum is laterally connected to the pelvic girdle. While all humans have the same number of cervical vertebrae, the number of vertebrae in other regions varies in about 5 % of the population (Marieb and Hoehn 2010). In an average adult, the vertebral column has a length of about 70 cm. From a lateral perspective, the spine describes an S shape; concave posteriorly in the cervical and lumbar sections and convex in the thoracic and sacral sections.

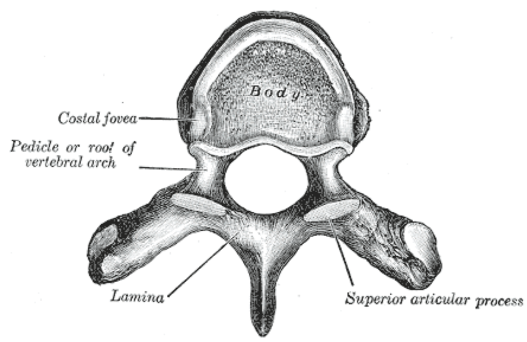

Figure 1.2: A typical thoracic vertebra from a superior perspective; the anterior side of the vertebra is in the top (adopted from Gray (2008))

Typical vertebrae (Figure 1.2) consist of an oval body at the anterior side, a vertebral arch to which the transverse processes, the articular facets, and the spinous process are attached at the posterior side, and the vertebral foramen, a hole in the vertebrae where the spinal cord is situated and which is mostly triangular for the cervical and lumbar vertebrae but rather circular for the thoracic vertebrae (Figure 1.2). Apart from between the atlas and the axis, all articulate vertebrae have intervertebral discs between them, consisting of an inner nucleus pulposus providing the disk with elasticity and compressibility and thus contributing to the spine range of motion, and the collagenous annulus fibrosus on the outside, providing stability to the disk.

2 AIMS

Accident statistics prove that pedestrian safety is still a major health issue in society world-wide, and that head injuries are the most fatal and disabling injuries in this area. Evaluation of pedestrian safety systems craves validated test tools that properly display human-like pedestrian kinematics. To date, the existing test tools such as human body models still require validation and improvement efforts in order to meet these goals to a satisfactory extent.

The ultimate goal is therefore to improve pedestrian protection with the intermediate goal to provide necessary human-like human body models to the industry. The main aim of this thesis is to gain a better understanding of head translational and rotational kinematics in pedestrian to vehicle accidents in the case of the arm and shoulder complex impacting the vehicle prior to head impact.

The goals of this work have been achieved through:

- Development of methods to quantify 3D translational and 3D rotational movement of spine, shoulder and head in full-scale pedestrian tests
- Creation of a data set suitable for use in pedestrian HBM evaluation
- Evaluation of an HBM in impacts to the shoulder, assessing shoulder deflection, thoracic and cervical spine motion and head movement of the model as in comparison to volunteers and PMHS
- Usage of HBMs to investigate how shoulder impacts with boundary conditions inspired by pedestrian accidents influence head motion

3 SUMMARY OF PAPERS

3.1 SUMMARY OF PAPER I

Data sets on full-scale pedestrian impacts suitable for validating HBMs are scarce. Validated test tools such as HBMs are necessary for the development and evaluation of pedestrian safety systems which are needed to address the persistent health issue resulting from pedestrian to passenger car crashes. One aim of Paper I was therefore to provide such validation data, including 3D translational and 3D rotational kinematics for the spine, the shoulder and the head. Another aim was to assess whether the specific loading conditions in these tests were similar to component tests and representative of real-life accidents.

A new full-scale PMHS experiment was conducted in order to provide 6 degrees of freedom (DOF) data and three PMHS tests from Schroeder et al. (1999) were re-analysed to provide times of impact of the elbow, shoulder, and head and to provide spine curvature through analysis of vertebral kinematics over time. The hands were not tied together in any of the experiments and different leg positions were used. In the new experiment, the subject was impacted laterally by a small size sedan car at a velocity of 40 km/h. The legs were positioned in a walking position where the impacted leg was slightly to the front and the contralateral leg was slightly to the back of the centre of gravity. The subject was instrumented with chess pattern photo targets on the head, T2, T11, L4, and both scapulae, and the impact was filmed with a high-speed video camera (Integrated Design Tools 2013). In the re-analysed experiments, one subject was impacted at 30 km/h and two subjects were impacted at 40 km/h while in a walking position with the impacted leg rearward. Accident data from the German In-Depth Accident Study (GIDAS) were analysed for occurrence of shoulder and elbow impacts in pedestrian accidents.

The experiments revealed that elbow impact had considerable influence on head kinematics, and that head kinematics prior to head impact were governed by neck kinematics. The spine was drawn downward shortly after first contact of the legs, inducing a slight roll backwards of the head. The lumbar and thoracic spine flexed away from the vehicle after pelvis impact. Subsequently, the torso rotated towards the vehicle while the head lagged behind the T1, resulting in neck flexion away from the vehicle. In the new experiment, the elbow then made contact with the vehicle and supported the thorax to some extent, while the head lag started to reduce shortly after elbow impact. In two out of the other three experiments, elbow impact did not appear to support the thorax as considerably. The head still lagged or the neck was just aligned with the thorax during impact in those tests. Elbow impact in the new experiment caused the shoulder to elevate rapidly before shoulder impact, causing the scapula to rotate upwards. Shoulder impact caused a reduction of the thorax relative to vehicle velocity which may increase the head relative to vehicle velocity as the neck flexion towards the vehicle increases. However, shoulder impact occurred only shortly before head impact in all experiments. As the duration of load transfer to the head played an important role, it was recommended to study the load transfer from the shoulder to the head in shoulder impact conditions similar to those in pedestrian accidents.

The study provides valuable head, shoulder, and spine 6DOF kinematics which can be used for HBM evaluation and detailed knowledge about the influence of upper arm and shoulder impacts on head kinematics in pedestrian accidents.

3.2 SUMMARY OF PAPER II

The aim of Paper II was to evaluate the suitability of the Total Human Model for Safety (THUMS) version 4.0 for studying head kinematics in pedestrian accidents with shoulder impacts.

Volunteer (Ono et al. 2005) and PMHS (Bolte et al. 2003) shoulder impact experiments were simulated with THUMS version 4. Head linear and angular displacements and vertebral linear displacements of the model were compared to volunteers and shoulder deflections to the volunteers and PMHSs. A parameter variation study was performed to assess the head response to a shoulder impact with varying shoulder posture and shoulder to vehicle impact conditions.

THUMS version 4 is generally suitable for studying head kinematics in pedestrian accidents with shoulder impacts but head rotation around the local z axis (head twist) requires improvement. Head linear displacements compare better to tensed than to relaxed volunteers and head rotation around the anteroposterior axis is comparable to volunteer responses. Head rotation around the lateral axis is small in volunteers and THUMS, and head twist is larger in THUMS than in the volunteers. The neck in THUMS is stiffer than in the relaxed volunteers.

The load transfer into the torso is controlled by the scapula motion over the thorax; the clavicle showed virtually no deformation. If the scapula cannot move relatively freely over the thorax, the shoulder response is rather stiff which generally leads to increased head linear and angular displacements. Raised shoulder postures stiffen the shoulder due to rotation of the scapula which brings the medial edge of the scapula closer to the spine, limiting its ability to move over the thorax. Head linear displacements were larger when the shoulder was raised. Changing the impactor surface orientation or the impact velocity angle, with regards to the impactor surface from lateral to supero-lateral made the scapula rotate downwards in addition to pushing it medially, and thus allowed for a somewhat less limited motion of the scapula. Head linear displacements were slightly decreased in these impacts. Distinct changes to the angle of the impacting surface and the impact velocity angle cause energy transfer reduction and a load path where the scapula moves less medially and rather rotates downward, thus reducing the shoulder coupling with the spine considerably.

In this study, about 30 – 40 ms passed from the first impactor to shoulder contact until the head began to move. The time span between shoulder impact and head first contact in pedestrian accidents was less than 20 ms in other studies. Thus, shoulder impacts will most likely not lead to head relative to vehicle speed reduction; however, they might contribute to head rebound velocity and thus possibly add to the overall velocity difference (delta v) of the head, immediately before and after head impact.

4 GENERAL DISCUSSION

This thesis investigates the influence of shoulder and spine response on head kinematics in pedestrian accidents. The catalyst for this investigation was the hypothesis that the overall stiffness of the shoulder would influence the amount of translational vs. rotational motion of the head before and during impact on the vehicle, and that the response of the spine would influence the load transfer to the head.

The main aim, gaining a better understanding of the head kinematics in pedestrian to passenger car accidents with shoulder impacts, has been pursued in two parts. Paper I investigated the linear and angular displacements of the head, shoulder and spine in full-scale pedestrian tests. In Paper II, the effect of shoulder impact on the head kinematics was studied more in depth. In this paper, shoulder impact experiments were used to evaluate THUMS version 4; then, the model was used to study head response to shoulder impacts with boundary conditions inspired by pedestrian impacts, where the shoulder posture, the angle of the surface that impacts the shoulder, and the impact velocity angle were varied.

For the present work, an FE model was chosen instead of a MB model. The author believes that injury prediction in pedestrian accidents will be more accurate by using tissue criteria which can be evaluated by an FE model. However, as mentioned in Chapter 1.5, FE models used in pedestrian accidents have to be evaluated in terms of kinematics before tissue criteria can be applied. Therefore, this work focusses on evaluation of the model kinematics rather than on tissue criteria evaluation.

4.1 METHODS AND ANALYSES USED IN THE PAPERS

PMHSs can be assumed to be the most accurate available surrogate for humans in pedestrian full-scale tests. However, subject preparation is a race against decay processes which may alter the subject response and reduce its biofidelity. In order to reduce this effect, embalming is sometimes used, although it is unknown how much embalming changes the subject's response. In the new experiment in Paper I, a combination of Winckler's embalming (Winckler 1974) and storage at 4 – 6 °C until the day of the test were used. The subject had extended joint stiffness, probably due to the effect of embalming, but the low temperature storage may also have contributed to this. The joints were thus moved prior to the experiment in order to remove the additional stiffness as much as possible. However, the force necessary to reach the full range of motion might have caused the bones to fracture; therefore it was not possible to remove the excess stiffness fully.

Pedestrian full-scale experiments with PMHSs comprise a number of inherent experimental difficulties, especially concerning subject positioning. The lack of muscle activity in the PMHSs leads to issues in positioning and a non-biofidelic lack of body stiffness in general. Several previous studies have highlighted that it was virtually impossible to bring the legs into a certain desired walking position (e.g., Subit et al. (2008), Kerrigan et al. (2005b)), which was attributed to lack of muscle activity. How to replicate realistic ground contact force has been in discussion since the beginning of pedestrian full-scale testing, but the initial solution of adding thin wooden sticks to aid maintaining posture (Pritz et al. 1975) has not been in wide use since then; as this practice is believed to stiffen up the leg response unnaturally. A final solution to PMHS positioning for pedestrian experiments has, to the best knowledge of the author, not yet been found. Different methods have been used in the past to hold the subject upright until the moment of impact. In Paper I, a dropping mechanism was installed, using a belt around the neck that attached to an electromagnet by metal strings; this appears to be a common method. Subit et al. (2008) placed a belt around the thorax under the shoulders instead. The dropping mechanism was, however, suspected to not reproduce a natural amount of force of the feet on the ground; instead, the belt produced forces where they would not be present in a real pedestrian accident, i.e., slightly extending the neck. The dropping mechanism thus limited the potential to change the overall posture of the upper body, i.e., spine curvature, shoulder posture, and head orientation.

Nevertheless, considering the level of violence involved in a typical pedestrian to vehicle, injury sustaining accident, and the variability of results stemming from variations in variables such as subject size, body proportions, or BLE-to-pelvis height, the author of this study expects that muscle activation plays a subordinate role in the overall kinematics. However, the author has not found any evidence of this issue having yet been studied in depth. Alvarez et al. (2013) show that neck muscle activation has some influence on head angular displacements in pedestrian accidents although the influence of leg stance on head angular rotations appears to be much greater. It is expected that HBMs will be able to significantly increase the understanding of the influence of active musculature on pedestrian kinematics.

Data collection in pedestrian experiments comprising PMHSs is a challenge in general, due to several reasons. Instrumentation is usually attached to the bones to enable direct measurement of the kinematics of the bones, a method also chosen in the experiment carried out for Paper I. In other experiments, photo targets have been attached to the skin or clothes, with the advantage of less tissue destruction, but at the expense of measuring the kinematics of skin or even clothes instead of the bones. If instrumentation is attached to the bones, it is generally virtually impossible to place them in the centre of gravity of the bone. Thus, data post-processing is needed to transfer the raw data into a coordinate system that can be easily understood, compared to other experiments, or be used for HBM validation.

Due to economic reasons, wired accelerometers were used in the new experiment in Paper I, which introduced cables from the subject to the acquisition system and thus added weight. In this experiment, the additional cable weight was less than 1 kg and was stored at the lower back of the subject. During the crash, the subject rotated onto its back and the author believes that this was predominantly due to lower extremity posture, where the impacted leg was forward. Since it cannot be excluded that the cable bag might have contributed to this rotation, it is recommended to distribute the cable weight on the contralateral side in the future.

In the new experiment in Paper I, a new method of video analysis was used which enabled acquisition of 3D translations and 3D rotations of each body part with a single camera. The video analysis software TEMA Automotive 6D (ImageSystems 2013) was used for lens calibration, as well as calculation of 3D translations and 3D successive Euler rotations. To allow for this calculation, special photo targets had to be used, consisting of a number of tracking points on a rigid surface. The coordinates of each tracking point relative to point of interest, i.e., a vertebral body CG, must be known so that this point of interest can be tracked. Using information from the lens calibration and inserting the real distances between the points on each target, the programme can calculate solutions to a set of 6DOF kinematics equations. If the number of points on each target is sufficient, this set of equations is over-determined and thus allows for error minimisation and improved accuracy. However, it has to be ensured that the coordinates of each of the tracking points are given in a correct coordinate system (and therefore their location needs to be measured very accurately), since this determines the coordinate system that the kinematics are calculated in. It appears that the accuracy of the video analysis with this system has the potential for significant improvement in comparison to other approaches of video analysis, where only one camera is used.

4.2 EXPERIMENTAL DATA FOR USE IN HBM EVALUATION

Evaluation of HBMs in pedestrian full-scale simulations is challenging due to the number of pedestrian experiments, which to date is strikingly low, considering the vast number of interdependent variables influencing pedestrian kinematics. Hardly any 6 DOF data on pedestrian kinematics is available, and even 3D translational data are scarce since most full-scale pedestrian studies present 2D linear kinematics. For evaluation of HBMs and for a better understanding of the human in-crash response, there is an urgent need for detailed experimental data produced in controlled full-scale pedestrian accidents. Paper I aims to provide such data for use in HBM evaluation for pedestrian accidents.

Another issue in HBM evaluation is the variability between the subjects, i.e., their size, weight, or body proportion. Numerous attempts have been made to scale experimental data in the past, often leading to reasonable results and a better understanding of the response curves. Scaling experimental data has the potential to be a valuable tool when validating HBMs, for instance, if the response of an average male for comparison with an HBM is of interest. However, scaling experimental data is not straight forward in pedestrian accidents. The pelvis-to-BLE height affects the upper body kinematics considerably, and the author has concluded that it cannot be compensated for by scaling experimental results. The author would thus recommend leaving the experimental data as is, allowing for the HBM to be compared to and evaluated against individual responses which the author considers would be a valuable contribution to HBM evaluation.

The test setup of the shoulder impact experiments is, compared to a full-scale pedestrian test, well defined and not subjected to the influence of a large number of variables as in a full-scale pedestrian test. For evaluation of the THUMS shoulder in Paper II, the volunteer experiment by Ono et al. (2005) and the PMHS experiment by Bolte et al. (2003) were used among the other available studies (see also Chapter 1.6.2). The main reasons were that these studies used a small impactor surface that did not load other body parts directly. In addition, the upper arms of the PMHSs were supported during the test. Where this was not the case (e.g., Bolte et al. (2000), Subit et al. (2010)), the shoulder was likely to be in a depressed condition, leading to a load path during impact that was different from the studies with volunteers and with PMHSs whose arms were supported. The quality of the data was deemed good in the case of Ono et al. (2005), where the boundary conditions were well-defined and the total number of subjects for each impact condition was 8 – 13 and thus fairly high. There was no visible oscillation of the VICON (Vicon Motion Systems 2013) targets on the volunteer skin during impact. The data quality in Bolte et al. (2003) was deemed acceptable. The number of subjects where data were available was relatively low at 3 – 4 for each load case. The oblique loading cases showed considerable seat back interaction and the distance between the upper back and the seat was not consistent among subjects; the oblique loading cases were therefore omitted from this part of the HBM evaluation. However, this experiment was still chosen above the other existing experiments since the elbows were supported in order to bring the shoulder into a neutral position, as discussed in Paper II.

Generally, as shown in Paper II, existing shoulder impact studies might not adequately mimic the shoulder impact in a pedestrian accident due to the angle of the impacting surface and the angle of the impact velocity. There is a strong need for additional experimental data for HBM validation so that response corridors can be established with adequate statistical significance. Thus, and even more so, each available data set provides highly valuable information that can, and should be, used to enhance available tools and knowledge.

4.3 SIMULATION RESULTS

As described in Chapter 4.2, before an HBM can be used, its biofidelity must be evaluated for a similar load case as the one it is intended to be used for. In the evaluation of the THUMS shoulder in Paper II, this was attempted by evaluating the model against two lateral shoulder impact experiments with additional oblique anterior and posterior impacts in one of the studies. Subsequently, the model was used in a similar way, based on the evaluation, but with different shoulder postures, angles of the impacting surface, and angles of the impact velocity.

In general, a decision has to be made at which point a model is biofidelic. Wismans et al. (2005) have established a definition of human body validation as “the process of assessing the reliability of a simulation model in comparison to one or more reference tests with human subjects”, where the experiments used for the validation should not be the same tests that were used for the original model input. According to Shigeta et al. (2009), none of the experiments used for the current work appear to have been used for model input. In addition, well-defined criteria of when the HBM can be regarded as “biofidelic” in a certain load case should be established before the assessment. This was not within the scope of the present work, but should be included in future HBM validation. Establishing such criteria for biofidelity could be done, i.e., by setting a limit of a certain percentage of the simulation results matching the experimental corridors, or by mathematical methods that compare the shapes of two curves. Several rating methods attempt to calculate such measures (e.g., Rhule et al. (2002), Sprague and Geers (2004), Hovenga et al. (2005), Gehre et al. (2009)) applying different approaches with varying results (Vavalle et al. 2013), underlining the difficulty in establishing one single rating method for biofidelity assessment of HBMs. Furthermore, the question remains how well the corridors display the average curves and spread of the living human population. While these are inherent limitations when discussing any attempts to validate HBMs, the validation process itself helps improving the understanding of the human body responses by continuing to bring forth new questions that can be addressed with new experimental setups.

The HBM is designed to simulate a male of around 177 cm stature, around 77 kg weight and adult age (around 30-40 years). In Ono et al. (2005), the average subject standing height, weight, and age were 171 cm ± 7 cm, 64 kg ± 12 kg, and 24 years ± 1 year, respectively. The standing height of the subjects in Bolte et al. (2003) was not given, but the average weight and age were 71 kg ± 12 kg and 74 years ± 8 years, respectively. Scaling can be used to reduce response variations between subjects and in an attempt to obtain the response of an average subject (Mertz 1984). When replicating the experimental data, both the data and the model were used without any scaling. Attempts were made to scale the experimental data from Ono et al. (2005) but an appropriate scaling factor could not be found. For shoulder deflections, shoulder width of the subjects appeared to be a good scaling candidate at first. In another attempt, the scaling method as described by Eppinger (1984) was applied. However, response variation among the subjects was not considerably reduced, which lead to both scaling attempts being discarded. Another candidate would be the effective mass of the impacted subjects. A higher effective mass would lead to a higher contact force and thus, generally greater shoulder deflections would be expected. However, as a general trend could not be observed when comparing maximum contact force and maximum shoulder deflection, this approach was discarded too. This indicates that other factors influence the shoulder deflections to a greater extent than the candidates named above. Such other factors may be the individual size of the scapula, its distance to the spine or its coupling to the thorax since the scapular motion over the thorax appears to govern the shoulder response, as discussed in Paper II. Scaling of the spine and head kinematics was not attempted since the individual spread was rather small in the unscaled data. Another approach would have been to scale the HBM to each subject's body proportions, although this would have contradicted the aim of evaluating the original HBM.

Information on positioning of the subjects in the experimental study was scarce, especially in Bolte et al. (2003). The model was positioned according to photo and video footage as accurately as possible, but since there were no measurements on the exact posture in the experimental studies, it is unknown how much the posture of the model actually deviates from those of the volunteers and the PMHSs. A study of the influence of posture was outside of the scope of this thesis. In the experimental data used for this work, it was expected that the model was positioned accurately enough for the related deviations to be of minor relevance. As discussed in Chapter 4.2, the oblique loading of Bolte et al. (2003) was discarded due to variations in upper back to seat back distances which appeared to influence the response considerably.

There are certain issues with data sets not originally designed for HBM validation since they are likely to lack important information for simulation input. In the data used for the present study, it was possible to reconstruct all necessary information from video footage and photographs to a satisfactory extent. In future experiments, ideally, all measurements required for simulation should be made available. Especially the subject initial position should be well documented, the experimental setup should be well defined, and all material properties should be measured for application to an existing material model.

4.4 INTERPRETATION OF RESULTS

In Paper I, it was found that elbow and shoulder impacts on the vehicle seem to occur rather frequently in full-scale pedestrian experiments, and that they both appear to have considerable influence on head kinematics. In Paper I, as well as in previous studies, see Chapter 1.6, the head was found to be lagging behind the torso until around the time of elbow impact, when it began to catch up. It was hypothesised that the elbow supported the thorax for a short period of time and slightly slowed down its motion towards the vehicle, giving the head some time to start catching up with the thorax. This hypothesis was supported by the video footage and by permanent bonnet deformation at the rear end of the bonnet, which could be attributed to elbow contact. However, there was no instrumentation installed to measure the loads transferred from the elbow impact into the torso.

The spine was found to be drawn downwards in relation to the surrounding tissue very early after first contact between the vehicle and the subject (Paper I). This might be an indication of how energy from the leg impact is transferred into the upper body at an early stage. It might also form part of an explanation for lumbar and thoracic spine injuries discovered in Subit et al. (2008). It is hypothesised that this mechanism is due to skeletal coupling between the long bones of the legs over the pelvic bones into the lower spine and then upwards up to the head: This is again supported by video footage where it can be seen how this particular dragging-downwards mechanism begins earlier in the lower vertebrae than in the upper ones.

In Paper II, it was found that THUMS 4.0 is a suitable tool for studying head linear and angular kinematics following shoulder impact in pedestrian accidents. However, the head twist was considerably higher in the model compared to the volunteers. A hypothesis would be that the model might not be able to properly compensate for the location of the head centre of gravity location in front and above the atlanto-occipital joint. The neck stiffness with respect to neck axial rotation should therefore be adjusted in THUMS version 4. The model compared better to the tensed than to the relaxed volunteers, indicating that a certain amount of muscle tension was added to the stiffness of the model during development. These observations should be taken into account if, in the future, THUMS will be fitted with active musculature.

The scapular motion over the thorax governs the shoulder response and the load transfer from the surface impacting the shoulder to the upper body in the model (Paper II). Motion of the scapula over the thorax was influenced by impact direction and shoulder posture. It can be expected that scapula size and geometry would play a role in different individuals as well. In the posterior impacts, the scapula was mainly pushed against the ribs with reduced ability to slide over the rib cage. Lateral and anterior impacts, in contrast, allowed the scapula to slide medially over the thorax, increasing shoulder deflections. Supero-lateral impacts did not produce significant variations of shoulder deflections as long as the scapular movement engaged the thorax and spine. However, when the angle of the impacting surface and the impact velocity angle became too large, the scapula moved and rotated mainly downwards, reducing the scapula to thorax and scapula to spine coupling considerably. With elevated shoulders, the medial edge of the scapula was rather close to the spine, which led to enhanced coupling between the scapula and the spine. Thus, it would be recommended to measure scapula geometry and orientation in neutral and elevated postures in future shoulder impact tests. In addition, it would be recommended to perform additional shoulder impact tests with elevated, as well as elevated and anteriorly displaced shoulders, as shoulder posture had considerable influence on head linear displacements as shown in Paper II.

The peak head velocity relative to the vehicle often occurs before head impact (Chapter 1.6). To date, thorough explanation has not been found for this phenomenon, to the best knowledge of the author. In addition to the findings in Paper I described in the previous paragraph, in Paper II, the way shoulder impacts influence head kinematics was investigated. One implication from Papers I and II is that elbow and shoulder impact might contribute to the head gaining speed in the direction of the vehicle path while at the same time limiting the downward motion of the upper body, thus reducing the head relative to vehicle velocity before head impact. This might provide part of the explanation for the peak head velocity occurring before head impact.

Nevertheless, the author believes that elbow impact contributes more to this effect than shoulder impact; the latter will not reduce head velocity significantly before head impact. It was discussed in Paper II that the time span from first contact between the impactor and the shoulder, until the head kinematics being affected, was about 30 – 40 ms. The time span in a pedestrian accident from first contact of the shoulder, to first contact of the head with the vehicle, was below 20 ms at vehicle speeds of 40 km/h in three experiments in Paper I and in two experiments in Kerrigan et al. (2007). However, the head resultant linear accelerations remained high for around 20 ms in Kerrigan et al. (2009) and the time span from head first contact to the deepest intrusion into the windshield in the new experiment in Paper I, was around 20 ms. This indicates that the head rebound might be affected by shoulder impact, thus increasing the overall head delta v (the velocity difference just before head first contact and just after head separation from the windshield), which could increase head injury risk. As pointed out by Watanabe et al. (2011), the speed of head and chest varied greatly after shoulder impact in a pedestrian impact against an SUV, which induces neck curvature and which might potentially increase neck injury risk.

Overall, it appears that shoulder impacts might increase the risk of injury to the head and neck in pedestrian accidents. In contrast, elbow impacts might reduce head injury risk since the time span between elbow impact and first contact of the head and the vehicle may be long enough to reduce the velocity difference between the head and the vehicle. In Paper I, the time spans between elbow and head first contacts was 56, 44, 38, and 37 ms, thus considerably longer than the time span between shoulder and head contact.

4.5 IMPLICATIONS FOR TESTING AND SAFETY SYSTEMS

Paper I and II indicate that the upper body kinematics, i.e., elbow and shoulder complex impacts, as well as spine kinematics, have considerable influence on head kinematics, i.e., head linear velocity and head rotation. As described in Chapter 1.4, current test protocols in Euro NCAP use only one impact speed (11.1 m/s) and one impact angle (65 ° to ground level for the adult head impactor) and do not measure head rotation. The author believes that head injury risk cannot be fully assessed with this method, as discussed in the sections below. The reasoning provided in this chapter is for adult. For children and shorter persons, lower pelvis to BLE height ratios may lead to different kinematics, and thus different conclusions.

As shown by other authors (e.g., Subit et al. (2008), Watanabe et al. (2012), or comparing Kerrigan et al. (2005a) and (2005b)), the upper body kinematics vary considerably with different vehicles used, while vehicle speeds were consistently around 11 m/s. As a result, the head impact velocities and the head impact angles with respect to the ground vary considerably among different vehicles. Resultant head velocities relative to the vehicle immediately before head impact were lowest in impacts with a SUV type of vehicle (Kerrigan et al. 2005b), around 7 – 9 m/s. In impacts with a medium sized sedan (Subit et al. 2008), a small city car, (Subit et al. 2008) or a small sedan (Kerrigan et al. 2005a), head impact velocities of around 11 – 14 m/s were measured. Head impact angles with respect to the ground level appeared to be similar in impacts with the SUV type of vehicle and the medium size sedan, nearly 90 °. In contrast, the head trajectory was much flatter in an impact with the small sedan or the small city car, where the head impact angle with respect to ground level was much lower and estimated at around 30 – 45 °. Head impactor testing should therefore at least take into account a larger range of head impact velocities and angles, ideally based on full-scale experiments or full-scale simulations with a similar vehicle front. In order to address brain injuries, it appears necessary to at least measure head rotational velocity and acceleration.

From the head impact velocities mentioned above, conclusion should not be drawn that SUV type of vehicles provide the lowest risk for fatal injuries in pedestrians. The pelvis kinematics presented in Kerrigan et al. (2005b) indicate that the pelvis might be at high risk of injury, followed by a presumably high load on the thorax during contact with the BLE. As mentioned in Chapter 1.1, thoracic injuries alone would have been the principal cause of fatalities in nearly 50 % of the fatal cases in Lau et al. (1998), although they may not appear in other statistics because they often coincide with fatal head injuries. Thus, the slightly reduced head impact speed observed in experiments with SUVs might, all things considered, not be as beneficial if it is accompanied by a considerably higher thoracic injury risk. Instead, the author believes that a sedan type of vehicle or a sports car would be most beneficial in terms of overall pedestrian injury risk. The front shapes of these types of vehicles appear to provide the longest time to head impact so that enough time might be available for the head to reduce the speed difference to the vehicle.

Furthermore, the present study has shown that impacts of the upper extremities, i.e., elbow and shoulder, have considerable influence on head kinematics. Elbow impacts appear to be one factor in reducing the velocity difference between the head and the vehicle before head impact. In contrast, shoulder impacts appear to occur too close in time to head first contact for them to reduce the velocity difference; instead, they might even increase injury risk as head rotational velocity appears to increase (Chapter 4.4) which has implications for the design of pedestrian safety systems. The author believes early contact of the elbow/shoulder complex to the vehicle would be beneficial and allow for a reduction of speed difference between the head and the vehicle prior to head impact.

5 CONCLUSIONS

The main aim of this thesis was to gain a better understanding of head translational and rotational kinematics in pedestrian to vehicle accidents in the case of the arm and shoulder complex impacting the vehicle prior to head impact.

The new pedestrian experiment in this thesis provides valuable 3D translational and 3D rotational displacements of the head, spine and scapulae which can be used when evaluating pedestrian HBMs. The effect of shoulder impacts on head kinematics was studied using THUMS version 4.

The main conclusions of this work are listed below:

- Head kinematics immediately prior to head impact are governed by neck lateral flexion.
- The lumbar and thoracic spine flexes away from the vehicle following pelvis impact. The neck flexes away from the vehicle shortly thereafter but neck flexion is reduced before head impact.
- Elbow impacts in pedestrian accidents have considerable influence on head kinematics. They may reduce head linear velocity before head impact, increase head angular velocity before and during head impact, and change the impact location on the head.
- Elbow impacts change the shoulder posture before shoulder impact to an elevated or elevated and anteriorly displaced position.
- Shoulder impact reduces the thorax relative to vehicle velocity.
- Shoulder impact occurs shortly before head impact and thus has limited influence on head to vehicle impact velocity and head rotational velocity before head impact; they might however contribute to head rebound velocity and head rotation during and after head impact.
- The scapular motion over the thorax controls the load transfer from shoulder impacts into the torso.
- If scapular motion is limited, e.g., through shoulder posture, the overall stiffness of the shoulder increases and leads to higher head linear and angular displacements after shoulder impacts.
- Elevated shoulder positions limit scapular motion over the thorax.
- Shoulder impacts from supero-lateral directions slightly increase scapular motion compared to lateral impacts.
- THUMS version 4 appears to be an appropriate tool to study head linear displacements following shoulder impacts.
- Head rotation around the anteroposterior axis is slightly lower in THUMS version 4 and THUMS displays considerably higher head twist than in the volunteers.
- THUMS version 4 head linear displacements compare better to tense than to relaxed volunteers.

6 FUTURE WORK

More full-scale pedestrian experiments which provide 6DOF kinematics of the head, spine, upper arms, and shoulders are required for evaluation of HBMs. Such experiments should provide detailed data on subject body proportions so that the influence of individual spread on the kinematics can be studied.

Full-scale HBM validation should be complemented through validation on component level. Specific body parts of volunteers or PMHSs should be impacted in a way that mimics the impacts in pedestrian full-scale experiments. Such experimental data is needed, especially for elbow, shoulder, and pelvis impacts.

It is important to evaluate the performance of the current average male HBMs against full-scale experiments with subjects of different sizes and proportions so that the industry gains more knowledge of how well the original model performs, considering the individual spread in real-life accidents.

Scaling the model individually should be considered since scaling of the experimental data is limited. Different HBM scaling methods should be evaluated, i.e., simple geometrical scaling to size and weight should be evaluated against morphing to individual proportions.

For full-scale pedestrian HBM evaluation, appropriate vehicle models, validated for pedestrian simulations, are required. Such validation can be achieved through impactor testing similar to the Euro NCAP pedestrian evaluation procedures. These tests should include impacts of a legform to the bumper, an upper leg impactor to the BLE, and headform to bonnet, windshield, the space between the bonnet and the windshield, the A-pillars and the upper windshield frame.

Regarding THUMS version 4, the differences in head rotations between THUMS and the volunteers indicate a need to address the neck properties in THUMS. Whilst THUMS already is a valuable tool for pedestrian safety, further improvements can be made.

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