The biomechanical aspects of pedestrian protection

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Abstract: In this paper a biomechanical basis for pedestrian protection is presented based on reviews of epidemiological and biomechanical studies conducted over the last three decades. Epidemiological studies reveal the nature and cause of pedestrian crashes and injuries sustained in the field. The various factors that influence pedestrian crashes and fatalities such as pedestrian demographics, time and location of crash, type of vehicles involved and their design characteristics, impact speeds, and nature and severity of injuries sustained are covered in the epidemiology section. The biomechanical studies identify the injury mechanisms and the biomechanical tolerances. Several biomechanical studies that attempt to identify the injury mechanisms and quantify the tolerances are critically reviewed in this paper, and the existing gaps in literature are identified. Further, the three primary injury mechanisms for pedestrian lower extremity injuries are highlighted, and an injury mechanism for depressed tibial fracture is hypothesized. The effect of exterior vehicle parameters such as bumper height, bumper stiffness, hood length, hood stiffness, bumper lead angle on the nature and severity of injuries sustained are also discussed. The biomechanical injury criteria and tolerance values in a proposed draft ECE pedestrian regulation are also presented. Finally conclusions are drawn based on the epidemiological and biomechanical studies, which lead to a proposal for future work.

Keywords: Biomechanics; crash; epidemiology; head; injury; knee; lateral impact; lower leg; pedestrian; safety.

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1 Introduction

The issue of pedestrian protection has led to a variety of epidemiological and biomechanical studies conducted over the last three decades. Significant progress has been made in understanding the nature of pedestrian crashes, injuries sustained, and the underlying injury mechanisms. The nature and cause of pedestrian crashes and injuries sustained in field are revealed by epidemiological studies. By conducting biomechanical investigations, the mechanisms of the injuries sustained and the biomechanical injury tolerances are identified. Biomechanical studies attempt to simulate the real world injuries in a controlled environment, and relate the mechanical input parameters (forces, moments, accelerations, etc.) to the resulting injury severity in terms of Abbreviated Injury Scale (AIS). The AIS numerical rating system ranges from zero (no injury) to six (maximum) and is a measure of the

mortality or threat to life (higher AIS level means increased threat to life). This paper covers both the epidemiological and biomechanical aspects of pedestrian protection.

2 Epidemiology

2.1 Crash statistics

In 1996, approximately 41,800 people were killed in the US due to traffic accidents. Pedestrian fatalities as a percentage of overall fatalities accounted for 11.3%, and passenger car and light truck occupants accounted for approx. 76.3%. Although the ratio of pedestrian accidents (85,000) to overall traffic accidents was only 1.2%, they accounted for 13.6% of fatal crashes (5090), and 3.3% of injurious crashes (75,000) [1]. Given the involvement rate of 1.2%, the number of fatalities is disproportionate. This implies an increased risk of fatality in a pedestrian crash relative to other crashes. One cause of this disproportionate risk is clearly the mass and stiffness difference between a vehicle and a person. Table 1 provides a summary of accident and fatality rates for the US. This increased severity may also be related to a number of factors – Blood Alcohol Level (BAC), weather, time of day, location of crash, speed of impacting vehicle, age of pedestrian, and the driver age among the major factors.

As noted in Table 1, almost half of the pedestrian fatalities are related to high blood alcohol levels, which may lead to a decrement in cognitive and sensory function, resulting in an increase in perception/reaction times. Additionally, research has shown that risk of injury is higher with alcohol in the blood than without.

Table 2 summarizes the pedestrian accident statistics in other countries. The data indicates that in the United Kingdom and Japan, pedestrian fatalities accounted for 26% and 28% of all traffic fatalities, respectively. In the Republic of Korea, pedestrians accounted for 37% of traffic deaths. Consequently, pedestrians in other countries seem to face a higher risk of fatality and injury from traffic accidents compared to those in US.

Table 1 US accident statistics [1]

<i>Type</i>	<i>Pedestrian</i>	<i>Total</i>	<i>% of total</i>
Fatal crashes	5,090	37,351	13.6
Injurious crashes	75,000	2,256,000	3.3
All crashes (fatal + injurious + prop. damage)	85,000	6,394,000	1.2
Persons killed	4,739	41,821	11.3
Fatality rate (per 100,000 population)	2.04	15.2	
Persons injured	78,000	3,189,000	2.4
Alcohol related deaths (BAC > 0.1 g/dl)	2,227	16,653	13.4
Percent	47%	40%	

Table 2 Accident statistics in other countries [2]

<i>Country</i>	<i>Fatalities</i>			
	<i>Total traffic</i>	<i>Pedestrian</i>	<i>Pedestrian as a % of total</i>	<i>Pedestrian per 100,000 population</i>
France	8,918	1,044	11.7	1.27
Germany	7,792	1,084	13.9	1.32
Italy	6,326	846	13.4	1.47
Japan	10,805	3,057	28.3	2.42
Korea (Rep. of)	10,416	3,874	37.2	8.42
Netherlands	1,066	110	10.3	0.70
Spain	5,957	995	16.7	2.53
Sweden	531	69	13.0	0.78
UK	3,581	946	26.4	1.60

2.2 Where and when pedestrian crashes occur?

The circumstances leading to pedestrian crashes have been hypothesized based on accident reconstruction, and driver/pedestrian/witness reports. The NASS database categorizes pedestrian crashes into two types: Intersection and Non-intersection crashes [1]. Table 3 shows the location of the crashes for the year 1996.

The NASS database also reports the factors related to pedestrian fatalities – 30.5% of them were due to walking, playing, working, etc., in roadway, while 30.5% were due to improper crossing of roadway or intersection, and 28% were due to darting/running into road or failure to yield right of way. The PCIDS reported that 86% of the pedestrians were attempting to cross the roadway prior to crash, suggesting that they were impacted on the side by the oncoming vehicle. Also, the pre-crash attention of the driver was noted as a factor. 60% of the drivers indicated that the vehicle was moving straight. During a turning maneuver, a pedestrian was more likely to be struck in a left turn. This was apparently due to the driver's attention focused on oncoming traffic while waiting for the traffic to clear, and failing to notice the pedestrian. In 68% of the cases, the driver's critical crash event was the pedestrian on the roadway, and the driver intended to continue straight through the intersection. This implies that the pedestrian did not have the right of way. About 43% of the drivers engaged in an avoidance maneuver, usually braking, and some steering [3,4]. Table 4 shows the occurrence of fatal/injurious crashes by

Table 3 Percentage of pedestrians killed/injured -- 1996

<i>Location of crash</i>	<i>Killed</i>	<i>Injured</i>
Intersection	23%	36.4%
Non-intersection	77%	60.7%

Table 4 **Percentage killed/injured by time of day**

<i>Time of day</i>	<i>Killed</i>	<i>Injured</i>
6 AM–6 PM	34.8%	63.7%
6 PM–midnight	47.6%	30%

time of day. Almost half of the fatalities occur from 6 PM–midnight, while 64% of the injurious crashes happen during the day especially, during 'peak hours' [NHTSA, 1996].

Most pedestrian crashes occur in city or residential streets where the population density is high [1, 3–6]. This is evident from the higher accident rates in Japan, the UK, and other European countries that have high level of urbanization.

2.3 Vehicles involved and characteristics

In the US during 1996, passenger cars were involved in 54% of fatal pedestrian crashes and 72% of injurious crashes, whereas light trucks were involved in 32% of fatal crashes and 25% of injurious crashes. The pedestrian was impacted by the front of the vehicle in 90% of the fatal crashes and 70% of the injury crashes [NHTSA, 1996]. The Pedestrian Crash Data Study (PCDS) analysis reported passenger car involvement at 71% and light trucks at 29%. This composition was proportional to the registration of cars and trucks in the US [3–6]. Out of a total of 2180 injuries for 292 pedestrians, 66% were from passenger car impacts, 15% from pick-up trucks, and 8% from SUV's. Further, 60% of the serious injuries (AIS3–AIS5) resulted from passenger cars, while pick-up's and SUV's contributed 22% and 6% respectively.

Based on the analysis of real world accident data reported by many researchers, pedestrian contact with different vehicle regions produced injuries to different body regions of varying severity (AIS 1–5). Children and adults sustained different injuries, though the vehicle contact regions were similar. Children sustained more injuries to the head, femur, and pelvis due to contact with the hood and hood-edge, while adults sustained lower leg and femur/pelvis injuries. Table 5 shows the distribution of vehicle contact areas and AIS, as reported in the PCDS. Almost 24% of the 2180 injuries reported were caused by contact with ground, and 6% due to unknown source [4]. From Table 5, it is evident that contact with the front bumper of the vehicle is associated with 20% of the injuries sustained by pedestrians. Of these, 76% are minor injuries (AIS 1–2), and 24% are serious injuries (AIS 3–5). Also, front bumper accounts for 20% of all minor injuries and 21% of all serious injuries sustained by pedestrians. Windshield is associated with a similar number of injuries. Other studies have reported similar findings [7–10]. In general, the front bumper is associated with lower leg/knee injuries in adults, and femur injuries in children. The hood surface is associated with head, femur, and chest injuries in adults, and head injuries in children. The likely head impact locations for children fall on the two-thirds of the hood surface closest to the hood edge, while for adults the location is closer to the windshield [8]. The hood edge is associated with femur/pelvis injuries in adults, and femur/pelvis/chest/abdominal injuries in children [7–9].

Table 5 Vehicle contact source vs. AIS [3,4] – total 1,524 injuries due to vehicle contact

Vehicle contact	Total (% of all vehicle contact injuries) [AIS 1–2, AIS 3–5]	% of all vehicle contact injuries	
		Minor (AIS 1–2)	Major (AIS 3–5)
Front bumper	20% [76%, 24%]	20%	21%
Hood edge	11% [71%, 29%]	10%	13%
Hood surface	14% [82.5%, 17.5%]	15%	10%
Reinforced hood	4.5% [66%, 34%]	4%	7%
Windshield	21% [74%, 26%]	20%	24%
A-pillar	5% [57%, 43%]	4%	10%
Front fender	4% [94%, 6%]	5%	1%
Cowl area	2% [72%, 28%]	1%	5%

The primary parameters reported in the literature that influence the vehicle-pedestrian kinematics and injury outcome are: bumper height from ground, hood edge height from ground, angle of hood, angle of windshield, length of hood, bumper lead length or bumper lead angle [4,11]. These are shown in Figure 1.

2.4 Impact speeds

Impact speed is one of the most difficult parameters to determine in a pedestrian accident investigation. The impact speed is one of the primary determinants of injury outcome of a crash. Several in-depth studies of accident investigations have tried to correlate impact speeds to injury in pedestrian accidents at speeds lower than 58 km/h. They also noted that 50% of all pedestrian fatalities occur at speeds less than 50 km/h, and there were no fatalities below 25 km/h. Figure 2 shows the relationship between impact speed and AIS as determined in the PCDS [4].

Figure 1 Vehicle parameters influencing pedestrian kinematics

Figure 2 Impact speed vs. AIS [4]

Ninety percent of AIS 1–2 injuries occur at impact speeds below 53 km/h. At an impact speed of 40 km/h and below, 80% of all AIS 1–2 injuries, 60% of all AIS 3–4 injuries, and 30% of all AIS 5–6 injuries occur. Figure 2 shows that the AIS 1–2 curve rises sharply with speed initially and then flattens out, demonstrating the high incidence of AIS 1–2 injuries at lower speeds. Given an impact speed, the variation in the AIS levels suggest that parameters other than impact speed also contribute to the injury severity.

2.5 Pedestrian demographics

Age is one of the most important factors in determining the severity outcome of a crash. It has been shown in the literature that children (0–15 years) and the elderly (>65 years) are disproportionately involved in fatal and serious injury crashes, compared to the rest of the population [1]. In an older study, EEVC reported the fatalities of pedestrians over 64 years old, as a percentage of all pedestrian fatalities: 35.4% – France; 47% – Germany; 40.4% – UK; and 48.6% – Italy [7]. An indication of the risk faced by pedestrians in a certain age group can be obtained by normalizing the fatalities in that age group to the population at risk (fatality rate per 100,000 persons). The pedestrian fatality rate was 1.2 for children, 2.0 for adults, and 3.5 for elderly in the US [NHTSA, 1996]. Compared to the rates for the adults in US, the rates are much higher in European countries during the 1970s: for the elderly – 15.6 in W. Germany; 16.0 in UK; and 11.1 in France [7]. The injury rates in US were 49 for children, and 18 for the elderly [NHTSA, 1996]. Based on the accident statistics for Europe, the elderly are exposed to the highest risk followed by the children.

Several in-depth field investigations of pedestrian accidents in US and Japan have been reported [3,4,6,12]. In Japan, 39% of all child traffic casualties, and 51% of all child traffic fatalities are due to pedestrian accidents [12]. From an analysis of 292 pedestrian crashes in the US, the involvement of children and elderly were 20% and 9% respectively [3,4]. Another study investigating 217 pedestrian fatalities in the US confirmed the increased fatality rate (9 per 100,000) for the elderly compared to other age groups (2 per 100,000) [6].

Pedestrian height is another important physical attribute in governing the dynamics of a crash. The severity of crash and type of injuries sustained depends on the relative height of the vehicle (bumper and hood edge) and the pedestrian (knee and hip height). A taller pedestrian may be more prone to head injuries due to contact with the windshield surround, while a shorter pedestrian may be at a higher risk for femur/pelvis injuries due to hood leading edge contact. Very few studies have reported the overall height of pedestrians. One such study is the Pedestrian Crash Data Study (PCDS) initiated by NHTSA in 1994. Of the 292 crashes reported, 19% of the pedestrians were 5 ft (60 in or 152 cm) or under, 36% were between 5 ft–5 ft 6 in (61–66 in or 155–168 cm), 28% were between 5 ft 7 in–6 ft (67–72 in or 170–183 cm), and 5% were taller than 6 ft (72 in [3,4]. Pedestrians between 5 ft–5 ft 6 in sustained mostly minor injuries (MAIS1–MAIS2), while those between 5 ft 7 in–6 ft sustained serious injuries (MAIS3–MAIS5).

2.6 *Body regions and severity of pedestrian injuries*

One of the problems encountered in the literature was inconsistency of injury scaling, some studies classified the injury as minor and non-minor, while others classified based on AIS-85 scale. Additionally, the definition of minor and non-minor injuries varied between studies, and the definition of body regions was also inconsistent.

Isenberg *et al.* indicated that 68% of all pedestrian crashes resulted in MAIS 1–2 level injuries, and 30% resulted in MAIS 3–6 level injuries. Considering injuries AIS 2 and above, the body regions most injured were head (32%), lower extremity (31%), upper extremity (9%), followed by abdomen, chest, and face (7% each) [3,4]. Another study reported that in fatal pedestrian accidents the most severely injured body regions were the head/neck/face (64%), the chest (12%), and the abdomen (8%). Among the non-fatal crashes, the body regions injured were the lower extremities (40%), the head (32%), the arm (12%), and the hip (10%). The same study found that the lower extremities (36%), the head (24%), and the chest (11%) were the most injured when considering AIS 2+ injuries [10].

3 Biomechanical aspects

3.1 *Overview of vehicle-pedestrian kinematics*

When a pedestrian is struck by the front of a car or a light truck/mini-van, the first contacts are usually with the bumper and/or the front edge of the hood depending on the shape of the vehicle front end. The region of contact on the pedestrian depends on the relative height of the pedestrian and vehicle front. For example, with small children the bumper may contact the upper leg and the leading edge of the hood may contact the chest. For adults the bumper may contact the lower leg/knee and the edge of the hood contact the femur/pelvis. The pedestrian may 'wrap-around' and slide over the front edge of the hood. This may allow the head and chest to contact the surface of hood/windshield. The location of this contact depends again on the relative height of the pedestrian and the geometry of the vehicle front end (length of hood, height of hood edge). At lower speeds (< 20 km/h), the second phase of impact – the

'wrap-around' or lateral bending of the pedestrian around the edge of hood, may not occur.

If the impact speeds are very high (> 60 km/h), the pedestrian may rotate about the second contact point (e.g. the head) and the whole body may contact the leading edge of the roof.

The forces resulting from contact with the vehicle front end accelerates the pedestrian up to the speed of the impacting vehicle. If there is braking of the vehicle, as the vehicle begins to decelerate, the pedestrian may move ahead of the vehicle and strike the ground [13]. This distance from the initial contact on the vehicle to the final impact point on the ground is commonly known as 'throw distance'. For example, at an impact speed of 40 km/h, the throw distance may be approximately 15 m.

The contact with the pedestrian may result in 'translation' and 'rotation' about anterior-posterior axis. The relative amount of each determines the pedestrian kinematics, which depend on the relative location of resultant contact force and CG of the body. If the impact is below the knee, the pedestrian may experience more rotation than translation. Also, as the speed increases, the likelihood of the head contact with the windshield also increases. Pedestrian kinematics is quite complex and depends on many factors such as impact speed, relative height of pedestrian and vehicle front-end geometry among others.

3.2 Injury mechanisms

To reduce the likelihood of pedestrian injury, it is helpful to understand the nature of the injuries sustained and the mechanisms of those injuries. Epidemiological studies have indicated the nature and severity of injuries – lower extremities and head are the most frequently injured body regions. Based on the Gadd Severity Index and the Wayne State Head Injury Tolerance Curve, a biomechanical head injury criterion for contact injuries – 'HIC' was established. An injury tolerance value of 1000 (for 36 ms duration) was established, and was used for many years in FMVSS 201 and 208 for head impact protection. Therefore, in the following sections, the discussion will be restricted to injury mechanisms of the lower extremity (lower leg, femur and knee), primarily during lateral impact. These injury mechanisms were investigated by the authors of the studies reviewed by conducting cadaver/dummy experiments in the lab, in an attempt to reproduce the injuries seen in field accidents.

3.2.1 Experimental studies

Experimental studies to investigate pedestrian injuries began in early 1960s and became more focused during the 1970s. Several full-scale experimental studies were carried out with whole cadavers and actual vehicle front structures. Kramer *et al.* [14] carried out 209 pendulum tests (4–8 m/s) with cadavers to assess the fracture tolerance of tibia. They reported infra-condylar comminuted fractures of the tibia, when the impact was to the proximal tibia. They also noted that there was a high frequency of tibial fractures for older cadavers due to osteoporosis. One of the first full-scale experimental simulations was conducted by Krieger *et al.* [15] using cadavers and 95th percentile anthropomorphic dummy at three speeds of 24, 32 and 40 km/h. The cadavers were impacted in a standing position with 80% of the body weight on the impacted leg. The linear and angular acceleration data measured in the

tests aided in understanding the gross pedestrian kinematics. The influence of vehicle front-end geometry and ground friction on pedestrian injuries was first investigated by Pritz *et al.* [16]. In a series of 15 experiments (10–30 mph), they reported comminuted supracondylar fractures of the femur, comminuted fracture of proximal tibia and separation of tibial intercondylar eminence. They did not report any ligament ruptures. They also observed depression of the lateral tibial plateau, which may have been induced by the lateral rotation of lower leg and femur about the lateral condyles (of knee) with hip and ankle as valgus pivots. This may presumably explain the large vertical ground reaction force (twice the body weight) they measured. They suggested that there is a transition in injury severity from 15 mph to 20 mph. Lowering the bumper led to reduction in injury severity to the lower leg, but increased the head impact velocity on the hood.

In another test series, Pritz [17] compared the responses and trajectories of the modified Part 572 dummies (50th percentile and 6-year-old child) with cadaver data. He reported that dummy neck and shoulder responses were too stiff compared to the cadaver. Eppinger and Pritz [18] were among the first to report injuries to the anterior (ACL) and posterior cruciate ligaments (PCL), combined with the tibial, fibular, and condylar fractures of the femur. From full-scale pedestrian tests, based on force-time data, they developed a leg-form impactor having the same effective mass, to simulate the leg interaction with the bumper. Aldman *et al.* [19] in a series of component tests using cadaver femurs and lower leg-ankle complexes with concentrated body mass illustrated the effect of impact location on tibia and corresponding bumper forces. They reported that the bending force at the knee was lowest when the impact was at the CG of the tibia. They also found that the simulated body mass had a minor influence on the bumper force and ground friction force except in low speed impacts.

Bunketorp *et al.* [20] simulated leg injuries from clinical studies in experiments using amputated femurs and whole legs (hip–femur–lower leg–ankle complex). Malleolar fractures and ligament injuries of the ankle joint were usually associated with bumper contacts at mid-tibia level. They hypothesized that these injuries resulted from shearing force of the distal tibia and the inertia of the accelerating foot. They also observed that at lower velocities, a lower bumper height resulted in undisplaced malleolar fractures of the ankle joint and ligament ruptures, while a higher bumper height resulted in medial collateral ligament (MCL) rupture of the knee joint. At higher velocities, a low bumper height resulted in comminuted fractures of the tibia and/or fibula, while a high bumper height resulted in fractures of the tibial and femoral condyles. It was noted that a flexed knee was more vulnerable to compression forces than an extended knee. They also suggested a combination of shear force, bending moment and compression force at the knee joint as an injury mechanism for the knee.

When the lower leg/knee is impacted on the lateral side, it may bend creating a compressive force on the lateral condyle of the tibia and a tensile force on the MCL and cruciate ligaments. This may result in compression fractures of the tibial plateau (also referred to as tibial condyle in literature) and ruptures of the ligaments [21]. A review of adult pedestrian injuries and their long-term prognosis was provided by Manoli [22]. He noted that knee injuries could be from valgus bending (due to impact on the lateral side) or varus bending (due to impact on medial side) of the knee.

Valgus stress may result in injury to MCL and medial meniscus, and may result in fracture of the lateral tibial plateau. They noted that intra-articular fractures and ligament ruptures of the knee joint could have serious long-term consequences.

Kallieris *et al.* [23] in a full-scale simulation with cadavers observed injuries to the spinal column (lumbar and cervical) in addition to wedge fractures of the mid-tibia and oblique/transverse fractures of the tibial head. He indicated that lumbar spine injuries were associated with 'wrapping' of the pedestrian around the vehicle's front-end, which resulted in bending between thoracic and lumbar vertebrae. The anatomic attachment of the ribs to the thoracic spine makes it less compliant than the lumbar spine. This may result in the maximum bending moment occurring in the lumbar region resulting in injury to the vertebral bodies and discs.

Further investigation to assess the influence of bumper height and impact speed on pedestrian injuries was conducted by Cesari *et al.* [24]. In a series of twenty cadaver tests, the bumper forces and knee angles were measured relative to the bumper height. They concluded that bumper force was not a good predictor of injury risk, since it did not show any correlation to bumper height or the nature of injuries. It can be postulated that the effective mass of the leg may not have changed significantly with impact location (for a given impact speed), so the bumper force was just proportional to the acceleration of the inertial mass, which did not vary much. They reported no knee joint injuries for bumper heights at, or below, a height corresponding to a level off the floor of 75% of the lower leg length (100% corresponded to knee level). Further, for impacts below knee level, they indicated that knee angle was more dependent on the bumper height than the impact speed. A 65% bumper height resulted in least amount of valgus rotation. In another series of cadaver experiments, Cesari *et al.* [25] reported that a low bumper height reduced the risk of ligament injuries, and higher impact speeds produced bone fractures rather than ligament injuries.

To understand the injury mechanism of the knee joint in pure lateral shear loading, Kajzer *et al.* [26] conducted low speed (15-20 km/h) tests using cadaver lower leg specimens. The test set-up is shown in Figure 3. The load cell near the knee measured the knee reaction force, which was indicative of the shear force through the knee joint. The load cell on the impactor measured the impact force.

Based on observations and analysis of test results, they proposed two injury mechanisms: First they noted that injury was related to the acceleration of the leg and occurred within 5 ms after impact. These injuries were classified as 'contact' injuries (fracture of the tibial lateral condyle, head of the fibula, and diaphysis fractures of the tibia and fibula). The second observation was related to the force transferred through the knee joint during impact as the thigh was accelerated. This injury occurred about 15–20 ms after impact, as indicated by the knee reaction force. This mechanism generated intra-articular injuries of the knee such as avulsions/rupture of ACL, MCL, LCL, fracture of tibial intercondylar eminence and femoral cartilage injury due to the contact between tibial intercondylar eminence and medial femoral condyle.

Kajzer *et al.* [27] extended this approach to determine the injury tolerance of the knee joint when subjected to low speed bending impacts. The legs were impacted at the ankle joint to induce maximum bending moment at the knee joint. As in the previous study, they observed a characteristic double peak for the bending moment

Figure 3 Test set-up for pure lateral shear [26]

at the knee joint. The first peak occurred within 5 ms, and was related to the failure of MCL, fracture of medial condyle and/or compression fracture of lateral condyle. The second peak occurred about 50 ms, and represented the maximal activation of non-damaged knee structures. This may have involved the stretching of ACL and PCL, but did not exceed their tolerance limits. Nevertheless, when the knee was subjected to lateral bending, the most frequent injury was partial or total rupture or avulsion of the MCL. In other full-scale tests (pedestrian-vehicle impacts) reported in the literature, the knee joint was loaded in bending and shear, the magnitude of each depending primarily on the location of contact with the tibia. In the previous study, the knee joint was in pure shear and therefore may not be representative of the actual loading condition experienced in crash tests. Also, the mean age of the cadaver specimens used was over 75 years of age.

To address some of the previous concerns, Kajzer *et al.* [28] conducted another series of high-speed shear tests, and bending tests at 40 km/h. They modified the test set-up for shear to induce a shear force and bending moment through the knee joint simultaneously, and used the same set-up as before for bending tests (Figure 4). The mean age of these cadaver specimens was 51 years.

In this study, the nature of injuries differed from the previous tests. Typical injuries in shear loading were avulsion of ACL, comminuted supracondylar and comminuted diaphysis fractures of the femur, and articular fracture of the tibial condyle. Figure 5 shows the typical injury patterns observed. The initial injuries observed in bending were avulsion of MCL/PCL/ACL, and comminuted supracondylar, transverse, oblique, or wedge fractures of the femur. (Figure 6). The difference in injuries observed between the previous tests and this series was attributed to impact speed. Low velocity impacts caused ligament ruptures, while high-speed impacts resulted in fractures and ligament injuries.

In shear tests, the damage mechanism (of articular fracture) was due to shearing and small bending deformation of the knee. This led to axial compressive forces

Figure 4 Test set-up for shear (a), and bending tests (b) [28]

between the femur and tibial condyles, resulting in articular fractures. The proposed mechanism is shown in Figure 7. The diaphysis and metaphysis fractures of the femur were attributed to the high stress concentration in the knee from a combination of shear force and bending moment in both the shear and bending tests.

When these experiments were repeated at lower speeds of 20 km/h [29], they concluded that both shear and bending tests caused mainly ligament injuries (ACL in

Figure 5 Typical injuries in shear tests [28]

Figure 6 Typical injuries in bending tests [28]

Figure 7 Injury mechanism of knee in shear loading [28]

shear loading and MCL in bending) in contrast with bone fractures observed in high-speed tests.

The experimental studies discussed above have investigated and reproduced injuries to the femur, knee, and tibia sustained by pedestrians in real world crashes. In summary, the injuries to the lower extremities have been explained by three proposed primary injury mechanisms:

- 1 Contact injuries due to acceleration of the leg (fracture of the femur and/or tibial shaft);
- 2 Injuries due to shear forces through the knee joint (ACL rupture/avulsion, tibial intercondylar eminence fracture, femoral cartilage injury);
- 3 Injuries due to bending moment at the knee joint (compression fracture of lateral femur condyle, tibial plateau fractures, and MCL rupture).

3.2.2 *Mathematical modeling*

Over the past three decades, a variety of mathematical models have been developed, and are still being developed as the knowledge base of different injuries and injury mechanisms is being expanded. This body of knowledge has led to the evolution of more complex models. If valid assumptions are made, and the vital anatomical characteristics and physical properties accounted for, math models have the potential to help uncover the underlying injury mechanisms, under different loading conditions.

Most of the mathematical models of pedestrian impact developed in the 1970s and 1980s were rigid body based models. They were essentially rigid bodies represented by planes and ellipsoids, and connected by a hinge or spherical joints [30–32]. More mathematically modeled joints (translational, planar, cylindrical and free) are now available to accurately represent the pedestrian kinematics in 3-D. The complexities of early pedestrian models vary from 3-segment to 15-segments, in 2-D and 3-D, to represent various body parts and capture the kinematics. Padgaonkar *et al.* [30] was among the first to develop a 15 segment 3-D model of the pedestrian, to understand the gross motion and forces involved in a pedestrian impact. In the past, model validation was based on trajectories and accelerations of body regions, which were measured in lab tests. Using these criteria it is interesting to note that an increase in complexity of the model from 7-segments to 15-segments did not lead to a proportional increase in accuracy [32,33].

It is worthwhile to mention that most of the modeling effort in the past has been to understand the effect of vehicle front-end designs on pedestrian kinematics and accelerations. Very little emphasis was placed on understanding the injury mechanisms, until the 1990s. Most recently, Yang and Kajzer [34] developed a 3-D MADYMO rigid-body model of the human knee. The knee joint model included the articular surfaces, ligaments, and capsule, represented by planes, ellipsoids and spring-damper elements. The anatomical coordinates for ligament-bone insertions were captured in the model. It was assumed that ligament strains of 12% and 20% corresponded to partial rupture and complete rupture, respectively. The ligament strains measured in the model correlated well with the experimental observations. The model was able to predict the lateral condyle contact force, which was higher

than the impact force. Based on relative elongations, the MCL had the greatest risk of injury, followed by ACL. Using this model they assessed the influence of upper body mass on the responses and found that, upper body mass influenced every parameter they measured, though the magnitudes were not reported. One of the drawbacks of this model was that the bone fractures could not be reproduced. Also, the ligaments, which are highly viscoelastic in reality, were represented by elastic elements. The previous model was improved by adding a fracture joint (at impact location) to simulate contact fractures [35]. The model performance improved, and the condyle contact forces reasonably correlated to condyle fractures observed in tests. Further improvements were made to the model, by incorporating the viscoelastic characteristics of the ligaments and representing the tibial intercondylar eminence in the geometry [36]. Based on the results of shear and bending simulations (carried out by Kajzer *et al.*), they confirmed some of the mechanisms of knee injury of their previous findings.

It is interesting to note that very few finite element (FE) models of the pedestrian lower limb exist in the literature. The first attempt to develop a full biomechanical model was made by Bermond *et al.* in 1994 [37]. This model included the tibia, femur, ACL, LCL, MCL, and PCL, and was designed to simulate pedestrian impacts using PAM-CRASH as the FE solver. The bones were modeled as deformable bodies using linear viscoelastic shell elements: 650 for the femur and 550 for the tibia. Ligaments were modeled as 1-D non-linear linkage elements. Bone geometry was obtained from scanned bone surfaces.

In 1996, Yang *et al.* [38] developed a more detailed finite element model of the lower limb including the tibia, femur, ACL, LCL, MCL, PCL, and a simplified foot for pedestrian impact analysis using DYNA3D as FE solver. The bones were modeled as deformable bodies using linear viscoelastic solid hexahedron elements: 3072 for the femur and 5364 for the tibia. Collateral ligaments were modeled with a combination of shell and spring elements, while cruciate ligaments were modeled with spring-damper elements. Bone geometry was simplified to reduce the modeling complexity and smooth the contact surfaces.

In 2000, Schuster *et al.* [39] and Takahashi *et al.* [40] independently developed two detailed finite element models of the pedestrian lower limb. Both of these models were based on bone surface data from Viewpoint Datalabs. Each was validated using both individual bone test data and the full lower limb studies conducted by Kajzer *et al.* These models represent the current state of the art of pedestrian lower limb simulation. No studies related to injury mechanisms have been reported from these FE models yet.

All pedestrian lower limb mathematical models suffer from a lack of biomechanical data. In particular, these models need both static and dynamic non-linear stress-strain curves for cortical and cancellous bone, tensile load-deflection (including rupture) properties for the ligaments under various strain rates, and non-linear compressive properties of surrounding muscles, menisci, and cartilage.

3.3 Injury tolerance and biomechanical response

To develop countermeasures for pedestrian protection, either through improvement in vehicle front-end designs or structural enhancements, it is useful to understand the

injury mechanisms and the tolerance for those injuries. A good understanding of injury mechanism aids in the development of a biomechanical injury criterion to predict the risk of injury when exposed to a crash. Knowledge of injury tolerance helps to establish Injury Assessment Reference Values (IARV). Further, to extend the understanding of injury tolerances and injury mechanisms to real world assessment of vehicle designs, it is necessary to either develop biofidelic dummies or human mathematical models. To achieve this, biomechanical response corridors have to be established for design and validation. The dummy responses should lie within the biomechanical corridors when impacted under similar test conditions.

Krieger *et al.* [15] also carried out full-scale experimental simulation with cadavers to document the three-dimensional kinematics of vehicle-pedestrian interaction. They measured linear and angular acceleration of the major body segments. Pritz *et al.* [16] in a series of 15 full-scale experiments at 10–30 mph reported a critical value of 45 g's for the pelvic fracture tolerance. They found no correlation between peak bumper forces and injury. In another study comparing the performance of modified Part 572 dummy with that of cadaver, Pritz [17] reported the acceleration-time histories of pelvis, left knee and left foot along with the two-dimensional trajectory of kinematics.

In an effort to understand the strain-rate response of ACL and MCL, Aldman *et al.* [19] conducted tensile tests to determine the ligament stiffness and tensile strength. They reported an ultimate strength of 1.1 kN at an elongation of 3 mm, and a stiffness of 370 N/mm for the ACL. The MCL stiffness doubled with increasing strain rate, from 70 N/mm (at 6 mm/s) to 150 N/mm (at 140 mm/s). Begeman *et al.* [41] also investigated the viscoelastic response of MCL by subjecting the extended knee to valgus strain. The total stretch varied between 5% to 15%, and the strain rate between 157%/s to 480%/s. Average failure load was about 2 kN. In another study evaluating the performance of prophylactic knee brace, France and Paulos [42] evaluated the response of knee ligaments to valgus loading at a low rate (76 mm/sec). The contribution of MCL, ACL, and PCL at the onset of MCL rupture was 80%, 11% and 9%. These results indicated that the MCL experienced tension immediately upon the application of lateral force. With an increase in valgus angulation (medial joint opening) the ACL and then the PCL began to offer resistance. Also, the tensile test results with MCL indicated that an increase in strain-rate (approx. 12 times) resulted in an increase in the ultimate load (28%) and stiffness (40%). To initiate MCL injury, a medial joint opening of 7 mm was required. To produce rupture 15 mm was required. Erickson *et al.* [43] in a dynamic evaluation of the prophylactic knee brace also confirmed the findings of France and Paulos [42].

In another related study, Yasuda *et al.* [44] measured the elongation of the ACL and MCL simultaneously during lateral impact to whole leg specimens. The findings are summarized as follows: peak elongation of MCL was 3.25% at zero° knee flexion and 4.25% at 30° flexion; peak elongation of the ACL averaged 1.8% at zero° flexion and 3.8% at 30° flexion. The time delay between the peak impact force and peak tension in MCL averaged 50 ms, and 61 ms in ACL. The maximum elongation time of MCL and ACL depended on the knee flexion angle. ACL elongation rate increased with the flexion angle. These results may help explain some of the underlying injury mechanisms of the knee joint subjected to a lateral load.

Fracture tolerance of the tibia has also been investigated. Kramer *et al.* [14] was among the first to perform a tolerance assessment of the tibia under dynamic loading in the anterior-posterior direction. Results from 209 cadaver tests showed a wide spread in fracture forces. The authors concluded that a force of 4.3 kN at an impact speed of 7.1 m/s, and 3.3 kN at 6.3 m/s produced a 50% likelihood of fractures. Bunketorp *et al.* [20] in an attempt to reproduce the clinical leg injuries with component tests in the lab using whole leg cadaver specimens, reported a fracture tolerance force of 1 kN or below for the tibia. He noted that high ground friction combined with low contact height increased the fracture risk when the cadaver was simulating a walking stance. Aldman [21] indicated that a bumper force of 4 kN was required to fracture an unloaded tibia, but 1 kN may be enough to fracture a tibia loaded by its body weight. EEVC noted that the lower leg tolerance mainly depended on the tibia, since fibula was a weak bone. They indicated a tolerance value ranging from 3.0 kN to 9.8 kN for the tibia (based on small surface impact), but did not rule out the possibility of higher tolerance to injury when impacted by bumper-like surface. A tolerance value of 200Nm was suggested for the bending moment. Nyquist *et al.* [45] conducted static and dynamic three point bending tests with twenty-two unembalmed human tibias. The loading was by a 25 mm rigid cylinder at the tibia midspan in the anterior-posterior and lateral-medial directions. He reported a tolerance value of 320 Nm and 280 Nm for males and females respectively.

Cesari *et al.* [24,25] conducted twenty cadaver pedestrian tests by varying the impact speed from 20 km/h to 39 km/h, and the bumper height from 60% (of the lower leg height) to 120%. They reported that leg injuries could occur at a peak bumper force of 2 kN or less, though the bumper force was not indicated to be a good predictor of risk. They observed no leg injuries when the knee bend angle was less than 15°. They indicated that a bumper force of 3.3 kN was associated with long bone fractures, while a force of 2 kN was not. Most recently Kajzer *et al.* [28,29] conducted tests with cadaver leg specimens to assess the injury tolerance of the knee to shearing and bending loads. The results of the tests are summarized below in Table 6.

Table 6 Results of shear and bending tests showing the initial damage mechanism [28,29]

Test type	Impact velocity = 20 km/h		Impact velocity = 40 km/h	
	Knee shear force (kN)	Knee bending mo- ment (Nm)	Knee shear force (kN)	Knee bending mo- ment (Nm)
Shear	2.4	418	2.6	489
	ACL injury initiated		Extra-articular fractures of the femur and tibia (supra-condylar or diaphysis fracture)	
Bending	1.3	307	1.5	331
	MCL injury initiated		Extra-articular fractures of the femur	

4 Discussion

Based on the studies conducted so far, several real world injuries to pedestrians have been reproduced in the lab and the injury mechanisms proposed. But, the injury mechanisms of depressed tibial plateau fractures, supracondylar fractures in high speed impacts, and longitudinal rotation of the pedestrian about the vertical axis are among the major issues that remain unknown.

Based on the knee joint anatomy and an understanding of biomechanical loading conditions, one hypothesis of an injury mechanism for depressed tibial plateau fracture can be proposed here. The lateral and medial condyles are two of the most important articulating surfaces in the knee. The average distance between the centers of contact of the femoral condyles at the tibia is between three to four inches. When the lower leg/knee is impacted laterally, there may be shear displacement of the tibia followed by a lateral rotation of the knee joint. This valgus rotation of the knee may occur about the lateral condyles, while the hip joint and talus may become the instantaneous center of rotation for the femur and tibia, respectively. This rotation can induce a compressive force, which may be proportional to the angular acceleration or angular velocity of the tibia. Essentially, the impact force on the lower leg may induce a vertical acceleration of the upper body and thigh (has to overcome the vertical resistance offered by the mass of upper body and thigh). Assuming a body mass of 160 lb, a vertical acceleration of 1–3 g's may result in a vertical force of 160–480 lbf (700 N–2100 N). This is well within the range of vertical loads reported in some studies. The injury mechanism is illustrated in Figure 8 schematically. Data for angular acceleration and angular velocity of the lower leg are not commonly reported.

Figure 8 Proposed injury mechanism for depressed tibial plateau fractures

Figure 9 Schematic of proposed EEVC regulatory sub-system tests

Next, the injury mechanism for supracondylar fracture of the femur (in Kajzer *et al.*, high-speed shear tests) may be related to the strain-rate effect of ligaments (mainly ACL). At high strain-rate, ACL stiffness may be high enough to stress the femoral condyle in shear resulting in fracture. The shear and bending tests conducted by Kajzer *et al.*, reported supracondylar fractures at high speed loading. It is noted that these fractures may also be related to the test set-up. It is possible that the plate measuring the knee reaction force may have caused these fractures, by edge loading. Instead of a rectangular plate, if a cylindrical shaped plate of large radius was used, the outcome may have been different.

The vertical rotation of the subject in the experimental tests has been reported, but no plausible explanation has been rendered so far. Lower limb FE model simulations reported by Schuster [46] may provide some insight to this phenomenon. When cross-car motion of the pedestrian was simulated at impact, the friction of the leg on the bumper may have induced a rotation of the lower limb. It is hypothesized that this initial rotation results in the overall body rotation observed in some tests. In a few cases with no cross-car pedestrian motion simulated, less pronounced lower limb rotations were also predicted to occur. These smaller rotations may be due to the fact that the leg is not an axi-symmetric body. More muscle on the posterior of the leg may result in greater force during the initial phase of an impact. This force may induce some rotation to the lower leg as the impact proceeds (Figure 9).

5 Conclusions

Based on the studies reviewed, the following conclusions are drawn:

- 1 Though the pedestrian crashes accounted for 1.2% of all traffic crashes in United States (1996), they accounted for 11.3% of all traffic fatalities, confirming the severity of these crashes. Blood alcohol level is an important factor in the fatal crashes with 48% of the pedestrian fatalities related to high BAC levels (> 0.1 g/dl).

- 2 The pedestrian injuries and fatalities were much higher in other countries such as France, Japan, UK, and Germany. The fatalities vary from 11%–37% of all traffic fatalities. Most pedestrian crashes occurred in cities and countries that have high population density and urbanization.
- 3 Based on the accident statistics of US and Europe, the elderly (> 65 years) and the children (< 15 years) were exposed to increased risk of fatalities and involvement in pedestrian crashes. The fatality/injury rates (per 100,000 persons) were high for both the age groups, compared to young to middle-aged adults (15–64 years).
- 4 Most pedestrian crashes do not take place at intersections. Improper crossing and failure to yield right of way were the main causes identified. Almost 50% of the pedestrian fatalities occurred between 6 PM–midnight, while 64% of the pedestrian injuries were sustained during the day (6 AM–6 PM).
- 5 Passenger cars had the highest involvement rate among vehicles (70%) in fatal as well as injurious crashes. In more than 80% of the pedestrian crashes, the pedestrian was struck on the side (lateral direction) by the oncoming vehicle.
- 6 Vehicle front bumper, windshield, hood surface, hood edge, and A-pillar were the main vehicle–pedestrian contact regions. The bumper was the main contact for lower leg injuries in adults, and femur/pelvis injuries in children. The hood edge was associated with femur/pelvis injuries in adults, and chest/abdomen/head injuries in children. The hood surface was associated with head/chest/arm injuries in adults, and head injuries in children. The windshield surround was the primary source of head injuries in adults.
- 7 The primary vehicle parameters that influenced the pedestrian kinematics and responses, were the bumper height, hood-edge height, bumper lead angle, length of hood, angle of hood and windshield. At a vehicle impact speed of upto 40 km/h: 80% of all AIS 1–2 injuries, 60% of all AIS 3–4 injuries, and 30% of all AIS 5–6 injuries are represented.
- 8 The body regions most injured were – the head, the legs, the arms, and the pelvis. When minor injuries (AIS 1–2) were considered, the legs followed by the head were the most frequently injured body regions, while the legs followed by the head were the most frequently injured body regions when non-minor injuries (AIS 3–5) were considered. Among the fatal injuries, the head/neck/face was the most severely injured body part followed by the chest, the abdomen and the pelvis. Injuries to the lower extremity were mostly non-fatal, may have long-term effects such as traumatic arthritis.
- 9 Children were at a higher risk for femur/chest/abdomen/head injuries, while the elderly had higher risk of pelvic fractures compared to adults. For a given crash severity, the elderly were at a higher risk of injury, than children and adults.

Based on the results of experimental studies, injuries to the lower extremities have been explained by three proposed injury mechanisms:

- Contact injuries due to acceleration of the leg (fracture of the femur and/or tibial shaft);
- Injuries associated with the shear force through the knee joint (ACL rupture avulsion, tibial intercondylar eminence fracture, femoral cartilage injury);

- Injuries associated with the bending moment at the knee joint (compression fracture of lateral femur condyle, tibial plateau fractures, and MCL rupture).

The injury tolerance data and biomechanical corridors can be improved by further biomechanical investigations, and conducting experiments in a controlled environment (for example, to understand load sharing between ligaments). A good understanding of the injury mechanisms and biomechanical tolerance will also enable the development of a biofidelic pedestrian dummy. The development of human math models can also be improved. The literature is sparse, and the huge benefits of math models are yet to be fully realized.

6 Future work

- Use the new finite element (FE) models of the lower extremity to understand and predict injury mechanisms under different loading conditions – for example, cartilage damage, extent of ligament rupture, and capsule stretch.
- The accident data from epidemiological studies are old. Further studies have to be conducted to assess the influence of changing vehicle designs, changing demographics, and changing composition of the fleet on the nature and frequency of injuries.
- It may be beneficial to determine the viscoelastic characteristics and load sharing between different ligaments of the knee, in high-speed lateral loading under different impact conditions. This will aid in further understanding of injury mechanisms, development and validation of mathematical/CAE models, and finally lead to improved biomechanical injury criteria.

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