FEMORAL IMPACT RESPONSE AND FRACTURE

David C. Viano
Biomedical Science Department
General Motors Research Laboratories
Warren, Michigan 48090
USA

Introduction

The response and tolerance of the femur to axial impact has been the subject of considerable interest in occupant protection research. This is due to the possibility of a wide range of lower extremity loadings during the management of occupant energy in an automobile accident. Continued research has provided new information in four major areas: 1) field accident injury [1-7], 2) mechanisms of femoral fracture [2,8], 3) femoral fracture tolerance [8-16], and 4) methods of laboratory evaluation of femoral fracture potential [17-27]. Although considerable research has been devoted to improving our understanding of the response of the lower extremity to impact loading, the large number of interdependent factors associated with the occurrence of lower extremity injury significantly complicates the study of injury mechanisms.

The purpose of this paper is to review the current frequency of field accident injury of the lower extremity in relation to other body injury and to the distribution within the lower extremity. Mechanisms of femoral fracture and tolerance information from cadaver tests are reviewed with particular emphasis on their relationship to current safety standards and methods for evaluating car occupant protection.

Field Accident Studies

The National Crash Severity Study [23] provides basic information on the severity of occupant injury according to body region and lesion type (Table 1, entries in each column total 100%). In our basic evaluation of the available accident statistics, skeletal fracture and soft tissue injury are cataloged separately by body region. The statistics are for car occupants receiving an injury of moderate or greater severity. It is interesting to note that approximately 20% of the AIS 2-4 injury appears to be to the lower extremity, where the primary lesion is skeletal fracture or dislocation. As such, injury to the lower extremity appears to be a major fraction of occupant injury in this range of injury severity. Occupant injury can be further ranked according to the relative frequency of fracture/dislocation injury (Table 2). Of the moderate to serious fracture/dislocation injury, the lower extremity ranks first for AIS 2-3 injury and second for AIS 4 injury from this data. Injury to the lower extremity also ranks high according to incidence by body region, and as such, appears to represent a major occurrence in non life-threatening occupant injury.

Table 1: Distribution of occupant crash injury (evaluation of NCSS Data [23])

Body Region	Lesion	System/Organ	Injury Severity (AIS) 2	3	4	5	6
Head	Fracture	Skeletal	11%	6%	15%	-	-
	Cont./Conc./Lac.	Nervous System	29%	3%	14%	35%	23%
Neck	Fracture	Skeletal	-	3%	-	3%	35%
	Lac./Disl.	Nervous System	-	-	-	-	9%
Thorax	Fracture	Skeletal	7%	23%	8%	-	-
	Cont./Lac.	CV/Resp.	-	5%	3%	21%	22%
Abdomen	Lac./Rupt.	Digestive	-	5%	12%	30%	-
	Lac./Rupt.	CV	-	-	19%	2%	-
Upper Extremity	Fx./Disl.	Skeletal	14%	10%	4%	-	-
Lower Extremity	Fx./Disl.	Skeletal	19%	24%	12%	-	-
Other	-	-	20%	21%	13%	9%	11%
% AIS $\geq$ 1			14.0%	9.0%	2.8%	2.0%	0.8%

Table 2: Rank order of fracture/dislocation and lesion frequency by body region and injury severity (adapted from Table 1)

	AIS 2		AIS 3		AIS 4	
Fracture/Dislocation Injury						
1	L. Ext.	19%	L. Ext.	24%	Head	15%
2	U. Ext.	14%	Thorax	23%	L. Ext.	12%
3	Head	11%	U. Ext.	10%	Thorax	8%
All Lesions						
1	Head	40%	Thorax	28%	Abd.	31%
2	L. Ext.	19%	L. Ext.	24%	Head	29%
3	U. Ext.	14%	U. Ext.	10%	L. Ext.	12%

Table 3: Accident investigations of lower extremity injury (LEI)

Accident Investigation	Collection Years	% Frontal	% Male	% Driver	# Occupants	# LEI	# LEI/Occ.	% LEI Most (Equal) Severe Injury	% LEI/ Injured Occ.
UCLA-TRG [4]	1962-1968	85%	74%	69%	186	375	2.02	--	46%
UM-MVMA/CPIR [5]	1965-1979	76%	63%	72%	175	340	1.94	44%	--
U8-DTEP/ARU [6]	1970-1978	All	76%	70%	101	123	1.22	--	39%
NCSS-2 [7]	1977-1978	62%	43%	69%	382	538	1.41	50%	38%
NCSS-3/4 [7]	1977-1978	71%	59%	64%	371	419	1.13	60%	23%

Table 4: Distribution of lower extremity injury

Accident Investigation	Body Region: Hip and Pelvis	Thigh	Knee	Leg	Ankle and Foot
RRL [2]	24.5%	32.4%	14.0%	19.6%	9.5%
UCLA-TRG [4]	14.0%	12.0%	52.0%	16.0%	6.0%
UM-MVMA/CPIR [5]	20.0%	14.0%	26.0%	21.0%	19.0%
U8-DTEP/ARU [6]	27.6%	27.6%	9.8%	14.7%	20.3%
NCSS-2 [7]	21.4%	15.4%	17.3%	16.5%	29.4%
NCSS-3/4 [7]	25.4%	23.4%	11.7%	22.5%	17.0%

An evaluation of accident investigation data available in the published literature of lower extremity injury (Table 3) indicates a significant reduction in the frequency of lower extremity injury. The number of lesions per incidence of lower extremity injury has been reduced from 2.0 in an early 1960 study [4], to a current level of 1.1 in the most recent evaluation of the NCSS injuries of AIS 3-4 severity [7]. This has similarly resulted in a reduction in the relative frequency of lower extremity injury for an injured occupant from an initial occurrence of 46% in the early UCLA study [4] to a current level of 23% [7]. Although it is difficult to assess the individual role of all the potential factors leading to such a significant reduction in the number and relative frequency of lower extremity injuries to occupants exposed to crash situations, several investigators [2,4,6,26] have indicated that design improvements have contributed significantly to a reduction in the frequency and severity of lower extremity injury. From the available field studies it is clear that approximately 70-80% of the injury to the lower extremity occurs in frontal accidents where approximately 60-70% of the front seat injured occupants are drivers. The studies also point out that injury of the lower extremity is frequently equal to or the most severe injury of the body for AIS 2-4 trauma when a lower extremity injury occurs, thus accounting for 50-60% of the most serious injury in this selection of current NCSS data [7].

The distribution of lower extremity injury according to the three major articulating joints and two long connecting bones (Table 4) indicates a rather uniform frequency of injury. Distribution of injury to the lower extremity provided in this table is on the basis of moderate to serious injury as recorded in various accident investigations during the last 20 years. However, the selection of accidents for inclusion in the investigations and the details of injury classification are different [see 2-7], and should be considered carefully in any interpretation of the accident statistics. In particular, the definition of injury by body region is quite subjective for each study, since a precise anatomical definition of the joints is not possible and must include some portion of long connecting bones, the femur and tibia/fibula. Thus, the actual division between the thigh-knee-leg body regions can lead to considerable variation in the overall reported frequency of injury to the various lower extremity regions. This is quite clear when the frequency of knee joint injury is reviewed for the various accident investigations. The occurrence of injury to the knee ranges from a low of 9.8% to a high of 52%. The majority of lower extremity injury (50-60%) is directed toward the knee-thigh-hip complex where direct fracture of the femur represents a minor fraction in the overall spectrum of lower extremity injury.

A detailed analysis of accident injury data collected by the University of Michigan (Table 5, entries in each column total 100%) shows the distribution of lower extremity injury by body region and type of lesion. It is obvious that a major fraction of the lower extremity injury is fracture, accounting for 75% of the AIS $\geq$ 2 lower extremity injury in this study. As a fraction of the total sample, fracture of the leg (tibia/fibula) accounts for 18.3% of the sample, followed by hip and pelvic fracture--15.6%, foot and ankle fracture--14.4%, knee joint fracture--13.8%, and fracture of the thigh--(femur) 12.6%. Clearly, from this sample, fracture of the femur is the least frequent lower extremity fracture lesion. Following fracture injury of the lower extremity bones in frequency of occurrence is laceration of the knee joint--9.1% of the sample and dislocation of the hip--2.4%. In this detailed accident investigation study, the articulating joint includes $\pm$ 5 cm of the long bones from

Table 5: Distribution of lower extremity injury of AIS $\geq$ 2 by type of lesion and body region (analysis of UM-MVMA/CPIR accident data [5])

Lesion		Hip and Pelvis	Thigh	Knee	Leg	Ankle and Foot
		Body Region				
Fracture	75%	78%	90%	53%	87%	76%
Laceration	12%	--	4%	35%	7%	4%
Dislocation	4%	12%	--	2%	--	6%
Cont./Abr.	4%	9%	4%	5%	3%	3%
Rupt./Avul.	3%	1%	2%	5%	3%	3%
Sprain	2%	--	--	--	--	8%
% AIS $\geq$ 2		20%	14%	26%	21%	19%

Table 6: Distribution of lower extremity injury by location of body lesion and principal impact direction (analysis of UM-MVMA/CPIR accident data [5])

Principal Impact Direction		Hip and Pelvis	Thigh	Knee	Leg	Ankle and Foot
		Body Region				
Frontal						
1	14%	15%	13%	18%	6%	15%
12	41%	30%	50%	45%	57%	56%
11	21%	6%	23%	23%	25%	19%
Oblique						
2	6%	9%	4%	5%	2%	--
10	6%	7%	4%	8%	6%	5%
Lateral						
3	4%	13%	2%	--	3%	2%
8	1%	6%	--	1%	--	2%
9	7%	15%	4%	--	--	2%
% AIS $\geq$ 2		20%	14%	26%	21%	19%

the joint center. Injury of the knee is the most frequent lower extremity injury--26% (14% fracture and 9% laceration), followed by the leg--21% (18% fracture) and hip/pelvis--20% (16% fracture and 2.4% dislocation) injury.

Injuries of the lower extremity are also shown distributed according to the location of lesion and the principal impact direction on the vehicle (Table 6, entries in each column total 100%). In the sample, 76% of the accidents are frontal and account for 80% of the lower extremity injury. It is interesting to note the rather even distribution of injury of the major body regions of the lower extremity for the frontal impacts, where the knee--22% and leg--18% are most frequently injured. However, the rather uniform distribution in injury frequency according to the major joints and long bones for primarily frontal impact is contrasted by a more specific distribution of injury in lateral impacts. Although lateral impact accounts for only 12% of the accidents and 10% of the lower extremity injury in the overall sample, it is associated with 34% of the AIS $\geq$ 2 hip/pelvis injury. More importantly, 70% of the lower extremity injury in the lateral impacts of this study is to the hip and pelvis.

Mechanisms of Femoral Fracture

A recent evaluation of the impact response and tolerance of the femur to axial load [16] provides basic new information on the underlying mechanisms of femoral fracture. The experimental methods of Melvin [13,14] were employed in a series of additional cadaver tests, where soft tissue covering the femur was denuded prior to impact. The evaluation of high-speed photographs of the impact kinematics and evaluation of femoral midshaft strains during rigid and padded knee loading allowed a study of the mechanisms of midshaft and condylar fracture. Interestingly, all the observed fracture could be linked to a site of tensile strain preceeding gross fracture of the femoral bone (Fig. 1). During knee impact the patella was observed to wedge the intercondylar eminence of the femoral condyles causing lateral displacement of the condyles. The wedging apparently produced a site of tensile strain on the surface of the femoral-patellar joint. In several experiments anterior bending of the femoral condyles produced tensile strain in the supercondylar region. At condylar fracture the shaft was seen to displace anteriorly relieving the stored strain energy.

Failure of the midshaft was specifically linked to tensile strain on the anterior surface of the midshaft. In particular, strain measurements at mid-shaft indicated a site of tensile strain of approximately 1.2% at the time of fracture. The tensile strain was a direct result of anteroposterior bending of the femoral shaft which produces tensile strains that exceed the uniform compression of the femur from the axial load at the knee joint. It is also postulated that failure of the femoral neck is a result of tensile strain on the posterior aspect of the femoral neck near the insertion of the head in the acetabelum, again at a site of tension.

Another important feature of the experiments duplicating Melvin's earlier impact work was that femoral kinematics and fracture were observed during impact loading. The actual level of force at fracture was recorded in this experimental situation. A major finding was the fact that the peak contact force occurred prior to fracture of the femur, such that fracture occurred at force levels that were 55-65% of the peak contact load (Table 7). Analysis of the rigid body mechanics of the femur during the early phase of loading indicates that a substantial portion of the applied contact load produces an inertial response of the femur, i.e., rigid body motions. There is a more gradual deformation response of the femur leading to fracture at a load below that of the peak contact force [16]. The results of this investigation provide additional insight into the reasons for the extremely high levels of contact force (18.5 ± 2.0 kN) reported by Melvin for experiments resulting in fracture of the femur. Since a considerable portion of the applied force is involved in accelerating the bony structure of the lower extremity during high velocity-short duration impacts, the recorded contact force is not an accurate measure of the force developing structural deformation and potential fracture of the femur.

Femoral Fracture Tolerance

A review of the static and dynamic tolerance of the femur under axial load (Table 8) includes information that has been reported previously. However, the peak loads included in this table for the study of Melvin [14] have been reduced by 35% to account for the significant fraction of contact load producing merely an inertial response (rigid body displacement) of the lower

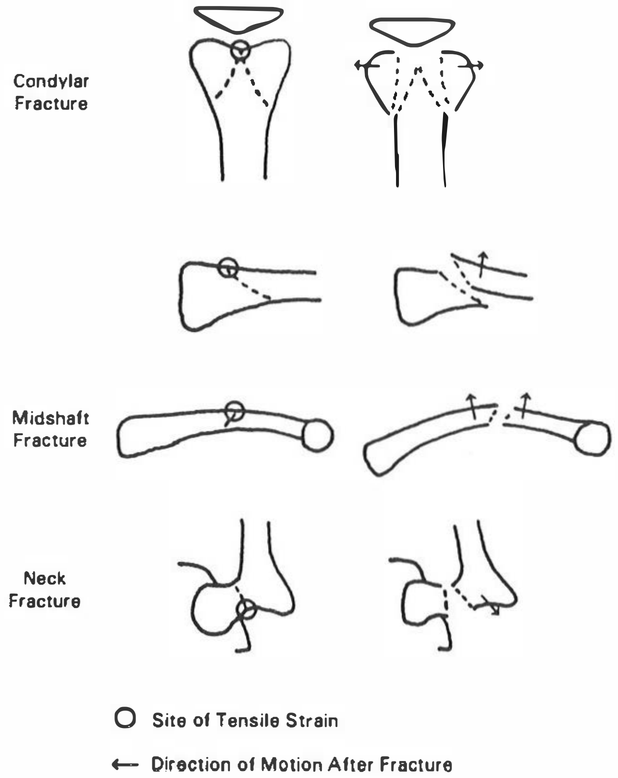

Fig. 1 - Tensile strain mechanism of femoral condyle, shaft and neck fracture resulting from axial impact (developed from [2,12, and 16]).

Table 7: Comparison of peak contact force and force at fracture for cadaver experiments using the methods of Melvin [13,14] (adapted from Viano and Stalnaker [16])

Impact Force	Interface Rigid (N=5)	 Padded (N=4)
Peak Value (kN)	16.3±5.4	12.4±3.9
Value at Fracture (kN)	10.6±2.7	6.8±1.0
Ratio	1.54	1.82

Table 8: Static and dynamic tolerance of the femur under axial load (adapted from [8])

Axial Loading	Fracture Load (kN)	 Duration (ms)	Nonfracture Load (kN)	 Duration (ms)
Static				
Carothers [9]	8.9±2.0	-	-	-
Fasola [10]	7.6-8.9	-	-	-
Dynamic				
Patrick [11]	8.9±2.0	26.1±3.4	8.0±3.4	25.0±4.8
Powell [12]	10.5±1.7	15.0±2.5	11.2±1.7	15.0±2.5
Melvin [14]	12.0*±1.3	8.1±4.9	9.1*±4.6	11.0±4.0

* Reported loads [8] reduced by 35% on the basis of results from [16], Table 7.

extremity. It is not suspected that a significant fraction of the contact load induces inertial acceleration in the experiments of Powell [12] due to the rigid support of the pelvis in his experimental methodology. In addition, the dynamic testing of Patrick is of a much longer duration loading such that the inertial component is expected to be a much smaller fraction of the reported contact load.

Although the static or long duration dynamic tolerance of the femur is based on rather limited data, it is approximately 8.9 kN [9-11]. The evaluation of factors leading to the extremely high forces recorded in Melvin's experiments provides additional emphasis to the need for realistic test methods. As reported earlier [8,15,16] the experiments of Melvin utilized a moderately lightweight impactor (10 kg) which had to be accelerated to relatively high contact velocities (10-20 m/s) to provide an impact of sufficient severity to potentially lead to fracture. However, in more recent laboratory simulations of knee restraint mechanics [19,20], it was emphasized that a contact velocity of approximately 6-8 m/s provides a more meaningful and realistic loading condition for the study of impact response and tolerance characteristics of the lower extremity.

Experimental data for the static and dynamic response of the femur to axial load indicates (Fig. 2) that the force to fracture the femur is dependent on the duration of the impact load. The femur responds statically for loads of duration greater than 20 ms; whereas, a structural dynamic response is stimulated for load durations below 20 ms. Higher contact loads are required to produce potential fracture of the femur when the load duration is substantially below 10 ms. Interestingly, the adjustment of the peak force recorded in Melvin's experiments brings the average fracture response into closer agreement with the predictions from a simple finite element model [15]. The model was developed to discuss potential fracture of the femur but does not address the potential effect of inertia on the femoral response. When strain rate effects are also considered [24,15], the load carrying ability of the femur is shown to further increase for short duration impacts. However, the nearly 10% increase in fracture load predicted on the basis of strain rate effects for a reduction from 40 ms to 10 ms duration of impact seems to be a secondary influence in comparison to the nearly 60% increase predicted on the basis of structural dynamics (see Fig. 2).

It is clear that peak loads on the order of 8.9 kN have a potential for fracture in human cadaver tests. However, the large variability in cadaver size, age, and structural integrity of the skeleton contributes to scatter in the experimental data for both fracture and nonfracture producing responses. The available data [9-14] and analyses [8,15,16,24] do provide a realistic basis for discussion of the fracture tolerance of the femur based on a wide range of experimental situations in various laboratories during the last 20 years.

Evaluation of Femoral Fracture Potential

Various laboratory simulations have been developed to evaluate the fracture potential due to lower extremity loadings. In particular, the development of the Hybrid test dummy has led to recording of femoral load as a component of the thigh structure. The reactive force can be measured in the dummy during impact. The Part 572 test dummy has been specified in federal safety standards as the surrogate for certification testing and the level of peak force developed in the dummy femur must be lower than 10.0 kN during barrier crash testing. The dummy has also been used in a series of laboratory simulations involving pendulum impact of the knee and sled testing to assess occupant dynamics during restraint system evaluations. Fortunately, many of the recent experiments [14,17,18] have involved comparative impacts where two surrogates, i.e., human cadavers and Hybrid dummy, were exposed to the same test situation.

One of the important results of these experiments is the fact that the level of contact force as measured in the human cadaver test is substantially different than that observed with the Part 572 dummy, or other available dummy surrogates. It is well recognized that the reaction force developed by the dummy is substantially greater (1.7-3.7 times [17]) than that of the similarly tested human cadaver.

Extensive evaluation of the comparative responses indicates that the massive rigid structure of the dummy leads to a much greater effective skeletal mass than the comparable human cadaver (i.e, 10 kg versus 2 kg) because the surface tissue and skin of the human cadaver is effectively uncoupled from the skeletal mass during impact [17,18]. As a result the dummy presents a greater mass during dynamic knee impact. The greater rigidity of the dummy knee also leads to more deformation of a compliant interface than the comparable human knee causing different interactive forces during deformation of the interface. In addition to the fact that the dummy contact forces are substantially greater than in comparable cadaver tests, it is also known that the femur load measured as a component of the dummy thigh is lower than the dynamic contact load because of the mass between the femur load cell and the knee joint in the dummy structure, i.e., the recorded femur load is approximately 70% [17,18] of the contact load during impact. When all factors are taken into consideration the ratio between the peak femur load in the Part 572 dummy and the contact load in comparable human cadaver (Table 9) indicates that the dummy femur load is approximately 50% greater than the human cadaver contact load. Therefore, the human cadaver responses reported in Table 8 and Figure 2 cannot be directly applied to the measured response of the Part 572 dummy.

Table 9: Ratio of peak force response of the Part 572 dummy femur load and human cadaver contact load (adapted from [8])

Test Condition	Force Ratio
Rigid Impactor [14]	1.20
Rigid Pendulum [17]	1.88
Padded Impactor [14]	1.43
Padded-Sled [18]	1.63
	1.54±0.29

Although it would be nice to suspect that a constant ratio exists between the dummy femur load and human cadaver contact load for many test situations, the wide range of interface conditions and the substantial dissimilarity between the Part 572 dummy and the human lower extremity structure does not lend itself to a uniform correspondence. On the basis of the range in ratios measured in various test situations, the current tolerance level used and specified in Federal Motor Vehicle Safety Standard (FMVSS) No. 208 occupant crash protection of 10.0 kN is not comparable to a single knee impact load in

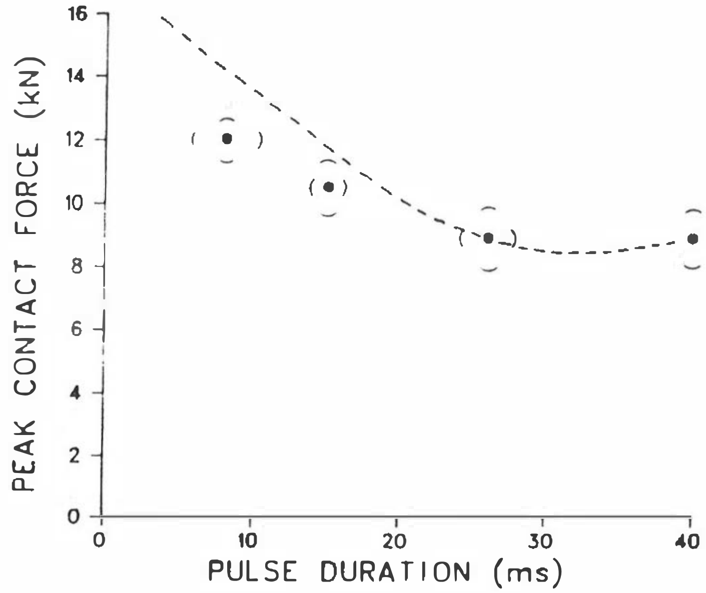

Fig. 2 - Dependence of femur fracture on the duration of impact loading due to a structural response for load durations less than 20 ms (modified from [8] by the results of [16]--dotted line is mathematical prediction of fracture [15] and mean data are plotted from Table 8).

a similarly tested human cadaver. Rather, a potentially wide range in impact load would be expected, depending on the particulars of the impact situation. However, a mean of 6.5 kN with a range of 5.3 kN to 8.3 kN would be predicted on the basis of the average data presented in Table 9. This range includes impact loads that are substantially below the average reported static fracture load in human cadaver tests. Since the range in ratio is very dependent on the test situation and impact interface, particularly emphasized in recent experiments [17,25], no conclusive comments can be made as to the comparative response between the Part 572 dummy and human cadaver in the general test environment.

Comparative tests with the Hybrid III dummy and human cadavers exposed to either a pendulum impact [19] or a sled simulation involving an experimental knee bolster [20] indicate that the dummy femur load is 1.11 ± 0.01 times that of the contact load in the human cadaver test. This ratio is substantially lower than that recorded in the laboratory tests with the Part 572 dummy reported in Table 9 but may be significantly affected by the energy absorbing interface material used in the well-controlled laboratory experiments, i.e., the ratio is expected to increase significantly as the rigidity of the interface increases and the duration of load decreases [17,25,8]. Clearly, the interpretation of data from a mechanical test device used in the evaluation of occupant protection requires care due to the variability in comparing responses with the human for all impact situations and interface.

Other mechanical procedures [21,22,26] have been developed to test the severity of knee impact without the use of an instrumented dummy. Lister and Wall [21] used a rigid knee-form (20 kg) which was accelerated as a pendulum into an instrument panel. The level of panel deformation was increased by raising the drop height. A peak knee-form force of 4-6 kN produced a level of instrument panel deformation that was equivalent to that observed in field accidents involving lower extremity injury. This type of laboratory simulation could then be used to provide meaningful information on the relative severity of knee impact with various interface structures, i.e., especially in A/B design comparisons. However, the difference in mass and stiffness between the knee-form and the human (effective mass of 2 kg during knee impact [17]) or the Part 572 dummy (10 kg [17]) makes a direct comparison of contact force to these surrogates difficult [27]. In fact, on the basis of simple analogies from previous work [17,18,8] or application of simple Hertzian contact of elastic bodies during impact, it would be expected that the human cadaver would require between 1.5-3.0 times the level of knee-form force to develop the same degree of instrument panel deformation. Additional comparative tests are needed to establish a preliminary correspondence between the contact force developed by the knee-form and either the Part 572 dummy or human cadaver.

Summary

The National Crash Severity Study (NCSS) and Motor Vehicle Manufacturers Association (MVMA) field accident studies are reviewed for vehicle occupants with lower extremity injury, especially femoral fracture. Interestingly, lower extremity injury is the most frequent fracture lesion of AIS 3 in the NCSS sample (lower extremity--24%, thorax--23%, upper extremity--10%, and head/neck--9%) and is a major factor for AIS 2-4 injury. However, femoral fracture is not the most frequent lower extremity injury. It accounts for only 14% of the AIS $\geq$ 2 lower extremity injury (knee--26%, leg--21%, hip/pelvis--20%, and ankle/foot--19%) in the University of Michigan--MVMA accident sample.

The higher than static load-carrying ability of the femur for short duration impact can be explained by a complex structural response which significantly attenuates local strain during dynamic loading. There is also a significant inertial acceleration component of the contact force which merely displaces the femur. Thus, the wide variation in dynamic load at femoral fracture observed in recent cadaver experiments can be explained by the geometry of the femur and the experimental impact methods employed. The available information points to a static tolerance of 8.9 ± 2.0 kN for femoral fracture, which increases for impacts of duration below 20 ms.

Femoral midshaft and condylar fracture has been analyzed in a series of axial knee impact experiments with a denuded cadaver femur. Fracture occurs primarily at a site of tensile strain even though the femur is subjected to compressive load at the knee joint. During compression the femoral shaft bends to produce tensile strain on the anterior surface of the midshaft. In addition, the patella wedges the condyles laterally and posteriorly, leading to potential tensile failure.

The evaluation of potential femoral fracture during axial loading has developed around the measurement of compressive femur force in a mechanical dummy. However, it has been shown that the contact load developed by the dummy significantly exceeds the comparable level in cadaver tests. The greater effective skeletal mass and rigidity of the dummy primarily contributes to the force difference, which is affected by the stiffness of the interface. The femoral load recorded in the dummy is also lower than the contact force due to inertial effects. Comparative tests indicate that the measured dummy femoral load averages 1.5 with a range of 1.2-1.9 times the contact load developed in human cadaver tests. Thus, a measured dummy femur force of 10.0 kN could be associated with a wide range (5.3-8.3 kN) in contact load on a human cadaver, depending on the impact situation.

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