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The Tolerance of the Human Hip to Dynamic Knee Loading

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ABSTRACT – Based on an analysis of the National Automotive Sampling System (NASS) database from calendar years 1995-2000, over 30,000 fractures and dislocations of the knee-thigh-hip (KTH) complex occur in frontal motor-vehicle crashes each year in the United States. This analysis also shows that the risk of hip injury is generally higher than the risks of knee and thigh injuries in frontal crashes, that hip injuries are occurring to adult occupants of all ages, and that most hip injuries occur at crash severities that are equal to, or less than, those used in FMVSS 208 and NCAP testing. Because previous biomechanical research produced mostly knee or distal femur injuries, and because knee and femur injuries were frequently documented in early crash investigation data, the femur has traditionally been viewed as the weakest part of the KTH complex. However, the relative risk of hip injuries to the risks of knee and thigh injuries in frontal crashes of late-model vehicles suggests that this may not be the case.

This study investigated the frontal-impact fracture tolerance of the hip in nineteen tests performed on the KTH complexes from sixteen unembalmed human cadavers. In each test, the pelvis was rigidly fixed by gripping the iliac wings with the thigh-to-pelvis angle set to correspond to a standard automotive-seated posture. A dynamic load was applied to the knee along the axis of the femur at loading rates that are representative of knee-to-knee bolster impacts in frontal crashes. Rigidly fixing the pelvis minimizes inertial effects along the KTH complex, which results in similar force levels along the KTH complex. Consequently, in these tests, the weakest part of the KTH complex failed first. All seventeen fixed pelvis tests that produced usable data resulted in acetabular fractures at an average applied force of 5.70 kN (sd = 1.38 kN). The lack of injuries to the femoral shaft and distal femur in these tests indicates that the tolerance of the hip is less than that of the femur under frontal-impact loading.

To further explore the tolerance of the femur relative to the hip, thirteen uninjured knee/femur specimens from seven cadavers previously used in hip tolerance tests were dynamically loaded. In these tests, the head of the femur was supported in a fixed "acetabular cup" to minimize inertial effects, and load was applied at the knee along the axis of the femur. All of these tests resulted in femoral neck fractures. Two tests also resulted in fractures to the femoral shaft. The average tolerance of the femoral neck from these tests is 7.59 kN (sd = 1.58 kN), which is significantly higher ($p < 0.05$) than the tolerance of the acetabulum. These results suggest that the mid and distal portions of the femur have a higher tolerance under these loading conditions than the pelvic and femoral portions of the hip.

KEYWORDS – Hip Fracture, Acetabular Fracture, Femur Fracture, Lower Extremities, Hip Tolerance, Femur Tolerance, Frontal Impacts.

INTRODUCTION

Lower Extremity Injuries in Early Biomechanical Testing

With the significant advances over the past decade in frontal crash protection for the head, neck, chest, and abdomen due to airbags and increased use of belt restraints, much attention in biomechanical research has been focused on disabling injuries to the lower extremities. The majority of this attention has been with regard to disabling injuries to the ankles and feet (Morgan et al. 1991, Pattimore et al. 1991, Crandall et al. 1996, Fildes et al. 1997). There has been considerably less attention given to injuries of the knee, thigh, and hip that result from the knees directly loading the knee bolster. In large part, this is because the response and injury of the knee-thigh-hip (KTH) complex due to direct loading of the flexed knee were studied extensively in the 1960s and 1970s (Patrick et al. 1966; Powell et al. 1974, 1975; Melvin et al. 1975, 1976). In these studies, tests were conducted by impacting the knees of whole cadavers with a rigid or lightly padded impactor, as well as by directing the knees into padded knee stops in unrestrained whole cadaver sled tests. These early KTH tests resulted primarily in patellar and distal femur fractures, but very few hip fractures.

The low incidence of hip fractures in early biomechanical studies was thought to indicate that the tolerance of the hip to knee impact is greater than the tolerance of either the femur or knee. This belief was supported by crash data from the same time period (before 1975), which showed that femur fractures occurred in 60% of frontal crashes that resulted in KTH injuries (Melvin et al. 1976). Also, because the femur was thought to be the weakest link in the KTH complex, protecting the femur was thought to protect the entire KTH complex (Viano 1977). Consequently, femur tolerance data from the dynamic knee impacts performed by Melvin, Patrick, and Powell were used to establish a maximum-force injury criterion of 10 kN for loading directed along the length of the femur. This tolerance was implemented as a KTH injury criterion in Federal Motor Vehicle Safety Standard (FMVSS) 208, which states that the force at mid femur of a midsize-male Hybrid III ATD shall not exceed 10 kN in vehicle compliance crash testing. The 10-kN tolerance has been correlated to a 35% risk of AIS 2+ injury to the KTH complex (Morgan et al. 1989).

Current Trends in KTH Injury Patterns

A recent analysis of real-world frontal-crash data in the National Automotive Sampling System (NASS) database from calendar years 1995-2000 shows that the incidence of AIS 2+ hip injuries in frontal crashes is higher than the incidence of knee or thigh injuries. In particular, the NASS data indicate that approximately 30,000 occupants sustain AIS 2+ injuries to the KTH complex annually in frontal crashes, and that approximately 47% of these are to the hip. For this analysis, the knee, thigh, and hip/pelvis are defined as illustrated in Figure 1, such that the knee includes the patella and femoral condyles, the thigh includes the supracondylar region, the shaft, and the subtrochanteric region of the femur and the hip/pelvis includes the femoral head and neck, and pelvis.

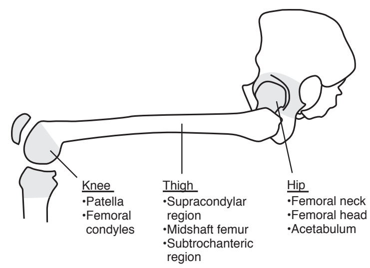

FIGURE 1. Definition of knee, thigh, and hip.

Figures 2 and 3 show the numbers and risks of knee, thigh, and hip injuries that are estimated to have occurred in frontal crashes by crash severity based on the weighted representative sampling in the 1995-2000 NASS database. In these plots, each risk value is calculated by dividing the number of occupants sustaining a particular type of injury by the number of occupants in the particular grouping that are estimated to have been exposed to frontal crashes. As expected, the risks of knee, thigh, and hip/pelvis injuries all increase with increasing crash severity, but the largest numbers of hip/pelvis injuries are occurring at crash severities that are equal to or less than current regulatory and consumer testing levels in FMVSS 208 and NCAP (i.e., less severe than 48-57 kph or 30-35 mph).

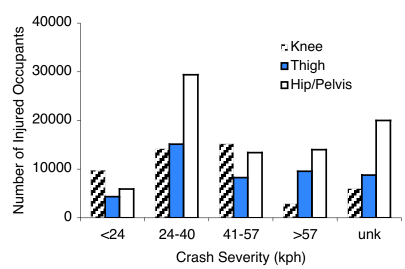

FIGURE 2. Number of occupants with AIS 2+ injuries in frontal crashes in NASS (1995-2000) by crash severity.

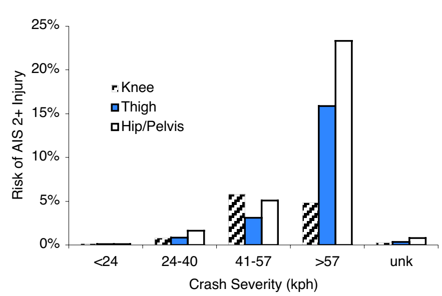

FIGURE 3. Risk of AIS 2+ KTH injuries in frontal crashes in NASS (1995-2000) by crash severity.

Figure 4 shows that the risk of sustaining a hip injury is greater than, or about equal to, the risks of sustaining knee or thigh injuries for all restraint conditions. Also, contrary to the expectation that older adults have lower fracture tolerance, Figures 5 and 6 show that there are no consistent trends in the risks of knee, thigh, and hip injuries with age, and that more occupants between 21 and 30 years sustain hip injuries than do adults in the older ten-year age groups because of the higher exposure of occupants in this age group to frontal crashes.

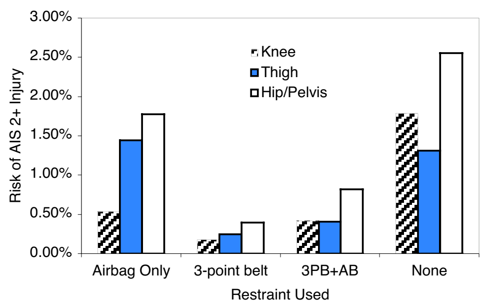

FIGURE 4. Risk of AIS 2+ knee, thigh, and hip injuries in frontal crashes in NASS (1995-2000) by restraint type.

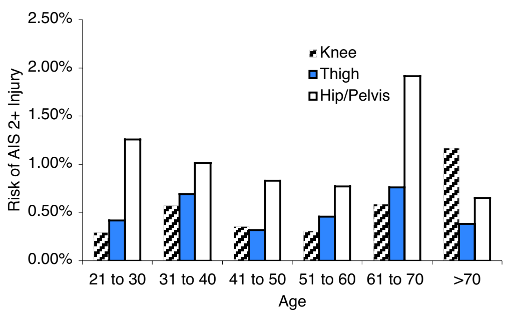

FIGURE 5. Risk of AIS 2+ knee, thigh, and hip/pelvis injuries in NASS (1995-2000) by occupant age.

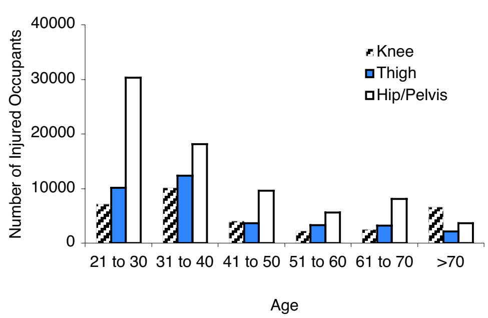

FIGURE 6. Numbers of occupants with AIS 2+ KTH injuries in frontal crashes in NASS (1995-2000) by occupant age.

Biomechanical Considerations for Hip Injuries

As indicated previously, early biomechanical research produced primarily knee or thigh injuries and rarely produced hip injuries. Therefore, it was thought that the tolerance of the hip was higher than that of the knee or femur.

One explanation for the relative inability of previous research to produce hip injury may be that previous tests used knee-loading rates that are higher than those experienced by occupants in crashes of newer-model vehicles. Figure 7 compares applied knee force-time curves from previous biomechanical studies to force-time curves obtained in frontal crash testing with late-model vehicles. The majority of the knee impacts in the studies used to develop the FMVSS 208 KTH injury criterion produced a fracture within 10 ms after the start of force application (Viano 1977), at loading rates between 400 and 3000 N/ms. However, typical force histories from a Hybrid III midsize male loading current model knee bolsters tend to peak between 20 and 60 ms at loading rates below 300 N/ms. While these differences in loading rate are not large enough to result in substantial differences in bone tolerance due to viscoelastic effects (McElhaney 1966), they may result in a time lag between the application of force to the knee and the onset of force at the hip, which could explain why early biomechanical studies rarely produced hip fracture.

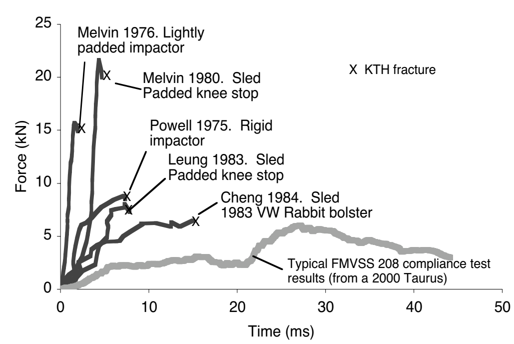

FIGURE 7. Knee force histories from existing studies relative to a typical FMVSS 208 femur load cell response.

More specifically, data from knee impacts performed on seated whole cadavers with denuded femurs by Stalnaker et al. (1977) and reported by Viano and Stalnaker (1980) indicate that there is a 2-to-5-ms time lag between the application of force at the knee and the development of a reaction force at the hip. Based on these data, at the high loading rates and resulting short times-to-fracture in the studies that form the basis for the 10-kN FMVSS 208 femur force criterion, it is hypothesized that the compliance of the

KTH complex resulted in time lags between the force applied at the knee and the resulting reaction force at the pelvis. These time lags are similar to times to fracture in the biomechanical studies used to develop FMVSS 208. Therefore, it is possible that previous biomechanical research failed to produce hip fracture because the tolerances of either the knee or femur were exceeded before force was transmitted to the hip, thereby preventing hip fracture. At the lower loading rates that are typical of knee bolster impacts in crashes involving current model vehicles, the hip may actually experience more force because the time required to generate sufficient force to fracture the knee or femur is much larger than the lag between force application at the knee and the resulting reaction force at the hip.

The mass distribution, mass coupling, and stiffness throughout the KTH complex can also affect how much force is transmitted to the hip from knee impact. Studies by Donnelly and Roberts (1987) and Horsch and Patrick (1976) have shown that impacts to the knee result in greater forces at the knee surface than at the mid shaft of the femur. These studies have shown that the decrease in force between the knee and mid shaft of the femur is proportional to the mass between the knee and mid femur. Since there is additional mass between the mid femur and the hip, the forces generated at the hip by knee loading can be assumed to be less than the force at the mid femur. This decrease is a function of the acceleration of the mass between the mid femur and the hip.

An analysis of hip/pelvis injuries in frontal crashes in the University of Michigan Crash Injury Research and Engineering Network (UM CIREN) database, shown in Figure 8, indicates a high incidence of acetabular fractures relative to other AIS 2+ hip/pelvis injuries. As expected based on the orientation of the pelvis relative to the direction of force applied through the femur, most acetabular fractures in the UM CIREN database are to the posterior acetabulum. Because case occupants in the UM CIREN database are selected from patients admitted to a level-1 trauma center, CIREN data are biased toward more severe crashes and injuries.

Figure 9 shows that acetabular fractures and other hip injuries (i.e., injuries to the femoral head and neck) in UM CIREN frontal crashes almost always occur on the side of the body toward which an occupant tends to laterally move. That is, occupants who moved forward and to the right (due to the location and/or angle of the frontal impact event) sustained primarily right-side hip injuries, and occupants who moved forward and to the left sustained primarily left-side hip injuries.

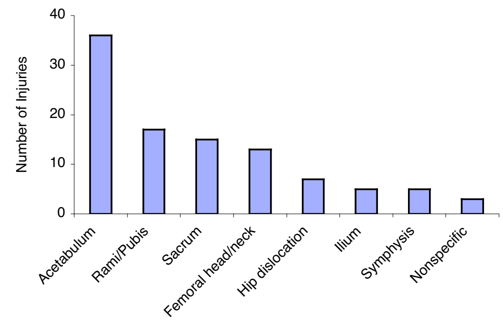

FIGURE 8. Distribution of hip/pelvis injuries in frontal crashes in the UM CIREN database (n = 101).

The relationship between direction of occupant motion and the side of the body that sustains hip injury suggests two possible factors that are contributing to the occurrence of hip fractures in frontal crashes. One factor is that occupant kinematics cause the right and left KTH to be differentially loaded. Specifically, the KTH complex on the side of the body that corresponds to the direction that the occupant would tend to move experiences higher loads than the contralateral KTH (e.g., the right KTH complex would experience higher loads if the occupant moves forward and to the right).

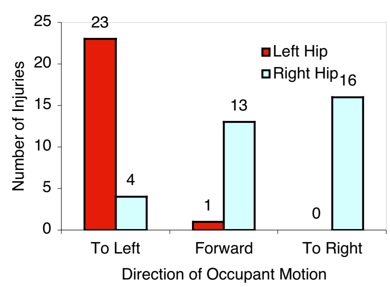

FIGURE 9. Distribution of hip (femoral head, femoral neck, and acetabulum) injuries by direction of occupant motion in frontal crashes in the UM CIREN database (n = 57).

A second factor is that changes in hip angle due to occupant kinematics in frontal crashes may act to reduce the tolerance of the hip. For example, in an angled frontal crash, the thigh on the side of the body toward the direction of body movement may tend to rotate inward toward the midline of the body after the knee contacts and loads the knee bolster. This adduction of the hip joint decreases the area of the acetabulum loaded by the femoral head and consequently decreases the tolerance of the hip (Letournel and Judet 1993). In addition, hip flexion caused by forward rotation of the occupant's torso and upward rotation of the thigh toward the torso, can further decrease the area of the acetabulum loaded by the head of the femur.

An analysis of cadaver knee impact data from previous studies (Patrick et al. 1966, Melvin et al. 1976 and 1980, Powell et al. 1975, Viano and Stalnaker 1980, Leung et al. 1983, Cheng et al. 1984, Donnelly and Roberts 1987, Yoganadan et al. 2001) supports the hypothesis that a change in femur orientation can increase the likelihood of hip fracture. The few hip fractures in these studies were generally produced when the orientation of the femur was changed so that flexion or adduction occurred either prior to or during impact loading.

Figure 9 also shows that occupants who moved straight forward sustained primarily right-side hip injuries. Since most of the occupants who sustained KTH injuries in the UM CIREN database are drivers, the prevalence of right-side hip injuries for these forward-moving kinematics suggests that braking is also a contributing factor to hip injuries, possibly by increasing the compressive forces on the hip due to muscle tension and by pressing on the brake pedal. Braking also may position the occupant's right knee so that it is in a more adducted posture when it contacts the knee-bolster area.

SIGNIFICANCE OF HIP INJURIES AND RESEARCH OBJECTIVES

The higher number and risk of hip injuries relative to the knee or thigh injuries in frontal crashes is of significant concern since hip injuries are clinically more severe and more difficult to treat than injuries to either the knee or thigh. Hip injuries often result in lifelong impaired gait (Nerubay et al. 1973) and account for the majority of life years lost due to KTH injuries in motor-vehicle crashes (Kuppa et al. 2001).

These real-world data and clinical implications of hip injuries, along with an analysis of the test conditions used in previous biomechanical studies conducted to determine response and tolerance of the KTH complex to loads applied at the knee, suggest a need for additional biomechanical research to better understand the response and tolerance of the KTH complex for knee loading conditions that are more representative of those resulting from frontal crashes of today's vehicles. To address these concerns and needs, a comprehensive research program is underway to study and quantify the factors that affect hip tolerance to knee loading in frontal impacts, and to develop a more comprehensive and accurate injury-prediction model for the complete knee, thigh, and hip complex under knee loading in frontal crashes.

This paper describes the methods and results of the initial phase of this research in which the tolerance of the hip has been quantified relative to the tolerance of the femur for thigh-to-pelvis geometries that are representative of adult occupants in a typical automotive-seated driving posture. In addition, the response and stiffness of the KTH complex have been determined for these loading conditions, and can be used to design and evaluate the biofidelity of the KTH complex of frontal crash dummies.

EXPERIMENTAL METHODS

Overview

In this study, hip injury tolerance in a standard seated posture was determined by dynamically loading the hip joints of unembalmed cadavers[1] along the axis of the femur. Loading rates used are representative of knee-to-knee-bolster loading in frontal crashes of newer model vehicles. Inertial effects were minimized by rigidly fixing the pelvis. This results in the same magnitude of force at the hip as at the knee or femur. Thus, for these test conditions, the weakest portion of the KTH will fail first.

To determine the tolerance of the femur relative to the hip, tests were conducted on the uninjured knee/femur complexes from a subset of the cadavers used to determine hip tolerance. Separately testing the hip and femur from the same subjects allows for direct comparisons between hip and femur tolerance that are independent of inter-subject variability in injury tolerance.

In this study, the terms femur and pelvis refer to all aspects of these anatomical components and are different from the knee, thigh, and hip regions, which are defined in Figure 1. The term femur includes the femoral condyles, shaft, head, and neck, and

[1] The rights, welfare, and informed consent of the subjects who participated in this study were observed under guidelines established by the U.S. Department of Heath and Human Services on Protection of Human Subjects and accomplished under medical research design protocol standards approved by the Committee to Review Grants for Clinical Research and Investigation Involving Human Beings, Medical School, The University of Michigan.

encompasses portions of the knee, thigh, and hip regions. Similarly, the pelvis includes the acetabular part of the hip, as well as other parts of the pelvis that are not considered part of the hip joint.

Hip Tolerance Tests

The test apparatus used to measure hip tolerance is illustrated in Figure 10. Prior to testing, the pelvis, sacrum, thighs, legs, and feet were removed as a unit from each unembalmed cadaver, and the flesh was removed from around the iliac wings, pubic rami, ischium, and proximal femur. To ensure that the hip joint capsule was not damaged, the flesh around the hip was not removed. The pelvis was then rigidly anchored to the test fixture by "clamping" the iliac wings to the iliac wing supports using clamps designed to distribute clamping forces and reduce stress concentrations in the ilium. These clamps were constructed by molding a polyester-potting compound to the medial and lateral contours of the iliac wings. As shown in Figure 11, the clamps and iliac wings were held together and to the test fixture using threaded rods and nuts. A support was also provided at the pubic symphysis to prevent pelvic rotation in the test fixture that could increase the likelihood of iliac-wing fracture near the clamps.

To load the KTH specimen at the knee, a pneumatic actuator was used to accelerate a weighted platform into a ram. The ram was connected to the knee of the cadaver by an interface fixture that was designed to prevent knee fractures by distributing forces over the patella and femoral condyles. The weighted platform was accelerated to a speed of approximately 1 m/s. A combination of Hexcel (3/8" cell diameter) and 13-mm-thick flotation foam was used at the interface between the ram and the platform to control the rate of loading and the magnitude of force applied to the KTH complex. Platform velocity, platform mass, and the character of the ram/platform interface were selected to generate loading rates at the knee below 300 N/ms and times to failure force between 20 and 60 ms. These are comparable to loading rates measured at the Hybrid III midsize male femur in FMVSS 208 compliance testing. For all tests, the combination of energy-absorbing materials that impacted the ram was similar.

A load cell attached to the ram just behind the molded knee interface measured force applied to the knee during each test. Measured force was inertially compensated using the ram acceleration and the mass between the ram and knee surface. Reaction force was measured at the pelvis by a load cell positioned behind the pelvis-mounting fixture. Based on analyses of the frequency content of the raw data, all forces and displacements were low-pass filtered using a 4th order Butterworth filter with a cutoff frequency of 300 Hz.

Prior to impact loading, the angle between the long axis of the femur and the plane defined by the ASIS (anterior-superior iliac spines) and the pubic symphysis was set to 120°. This corresponds to the hip flexion angle in a standard automotive-seated posture defined by Schneider et al. (1983). The upper half of Figure 12 shows a side view of the pelvis and femur orientation in this posture. As is illustrated in the lower half of Figure 12, the femur abduction/adduction angle was set so that a line connecting the mid point of the femoral condyles to the hip joint center was perpendicular to a line connecting the left and right hip joint centers, where hip joint center was estimated by palpating for the head of the femur. The combination of the flexion and adduction angles defined in Figure 12 is called the "neutral" posture.

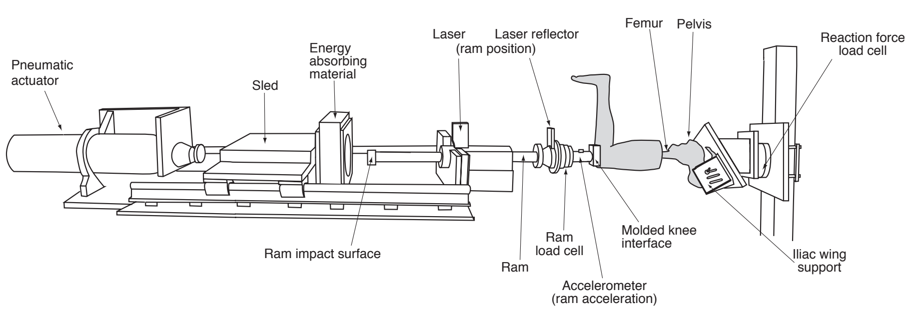

FIGURE 10. Apparatus used for hip and femur tolerance testing.

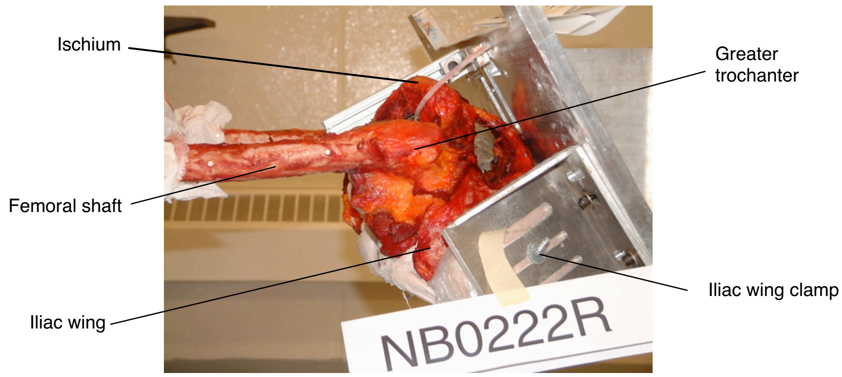

FIGURE 11. Side view of inverted cadaver pelvis mounted in test fixture.

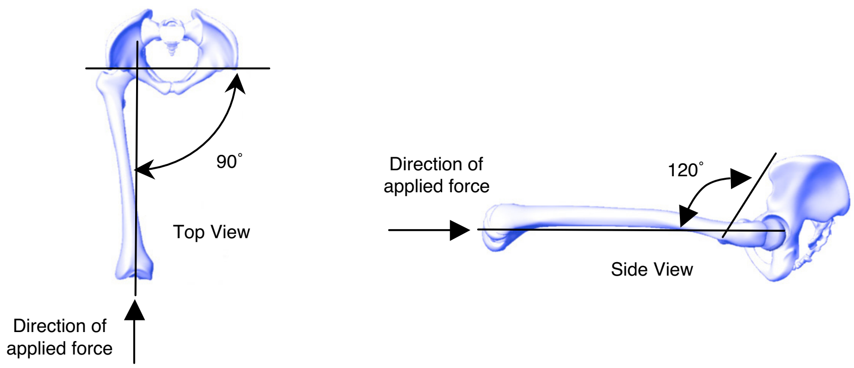

FIGURE 12. Thigh adduction (top) and flexion (bottom) angles used in the hip tolerance tests.

In the first three tests, sections of the lower extremities distal to the mid shaft of the femur were removed from the cadaver for other uses prior to hip tolerance testing. For these tests, an interface fixture was molded to the truncated end femur and the ram was positioned so that it initially contacted this fixture. Force was always applied along the axis defined by the femoral condyles and hip joint center. For tests where a knee was not present, the line of force application was offset from the centerline of the femur so that force was still applied along the axis defined by the femoral condyles and hip joint center.

Fracture force, time-to-fracture, and loading rate were determined from the applied force histories, as illustrated in Figure 13. Based on high-speed video of the test specimen during impact loading, fracture force was always considered to be the first peak in the force history, so that time-to-fracture force is the same as time-to-peak force. Time-to-fracture is the time required for the force to rise from 500 N to the first peak in the force curve. Loading rate was calculated as the slope of the force curve, based on a least-squares fitting of a straight line connecting 15% and 85% of the first peak in the force curve. The axial stiffness of the test specimen was calculated from the applied force and ram displacement in the same manner that loading rate was calculated from the force histories.

Following each test, x-rays of the entire KTH complex were taken. An autopsy was then performed to document injuries to the hip/pelvis.

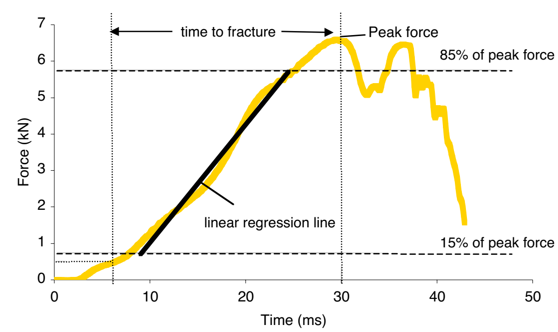

FIGURE 13. Typical applied force history showing loading rate, time-to-fracture, and peak force.

Femur Tolerance Tests

Following hip tolerance testing, the hips of some specimens were disarticulated and the left and right knee/femur complexes were separated from the pelvis for use in femur tolerance testing. These tests were also performed using the test device shown in Figure 10, but with the pelvis-mounting fixture replaced by a hemispherical "acetabular" cup to capture the head of the femur and transfer load from the femoral head to the reaction force load cell. A thin sheet of rubber was placed between the femoral head and the acetabular cup to reduce the stress concentrations on the femoral head. Loading conditions for the femur tolerance tests were similar to those used in the hip tolerance tests. Load was applied by the ram through a molded knee interface to reduce the likelihood of knee injury by distributing the applied load over the femoral condyles and patella.

Test Subjects and Test Numbers

Table 1 provides information on the seventeen cadavers from which specimens were obtained for testing and indicates the types of tests conducted on each specimen. Both the left and right KTH complexes from these subjects were used in hip tolerance testing. However, some tests on these specimens were conducted at hip postures other than neutral and will be reported in a future paper when testing for hip posture effects is complete. Table 1 contains bone mineral densities (BMD) and *t-scores* for the subjects when available. These data were obtained using the osteogram method, which uses a calibrated x-ray of the phalanges to determine a BMD^2 index. The BMD^2 index is used to calculate the *t*-score, which is bone density normalized with respect to a 25- to 50-year-old reference population of the same gender. Osteogram data are comparable to dual x-ray absorptiometry (Yang et al. 1994).

Test IDs beginning with the same number were conducted on left and right KTH segments from the same cadaver. All test IDs ending with L and R indicate left and right hip tolerance tests, respectively. All test numbers ending in RF and LF indicate right and left femur tolerance tests, respectively.

Six female and ten male subjects were used in nineteen hip tolerance tests. Tests 5L and 5R, 8L and 8R, and 25L and 25R are identical tests conducted on the left and right sides of the same cadaver to determine if a pre-existing contralateral fracture affects the tolerance of the hip (e.g., does a preexisting fracture of the left hip affect the tolerance of the right hip).

Two female and five male subjects were used in thirteen fixed knee/femur tests. All of these subjects were also used in hip tolerance tests. However, for Subject 14, valid data in the neutral posture were not obtained, and therefore, hip tolerance data from this subject are not reported.

TABLE 1. Test Subjects and Test Identifications

Subject Number	Gender	Age	Stature (cm)	Mass (kg)	*t*-score	BMD^2 Index	Test IDs: Hip Tests Left	Hip Tests Right	Femur Tests Left	Femur Tests Right
1	F	55	163	113	NA[†]	NA	5L	5R		
2	M	86	173	91	NA	NA	6L			
3	F	51	168	78	-1.3	96.3		7R		
4	M	79	180	82	-0.6	80.3	8L	8R		
5	M	60	178	125	-0.1	98.9	10L			
6	M	72	173	81	0.4	96.7		12R		
7	F	66	147	99	-1.4	106.9		13R		
8	F	68	165	71	-1.2	97.6		14R		
9	F	71	178	82	NA	NA		16R		
10	M	75	175	72	0.4	93.6	17L			
11	M	72	178	82	-0.5	86.6	18L			18RF
12	M	75	180	81	0.1	90.9		19R	19LF	19RF
13	M	41	176	91	0.7	118.7	22L		22LF	22RF
14	F	79	152	59	1.5	126.9			23LF	23RF
15	M	60	178	82	-0.9	89.4		24R	24LF	24RF
16	F	86	168	68	-1.6	93.5	25L	25R	25LF	25RF
17	M	62	183	91	0.1	99.8	26L		26LF	26RF
	Mean	*68*	*171*	*84*	*-0.31*					
	sd	*15*	*10*	*16*	*0.91*					

[†]NA = Not Available

RESULTS

Failure Loads and Injuries in Hip Tolerance Tests

Table 2 lists the results of the nineteen hip tolerance tests conducted in the neutral hip posture. Two of these tests failed to produce usable data. Test 7R, failed to produce injury, and test 13R resulted in an iliac wing fracture that was attributed to the mounting methods. All seventeen of the remaining tests resulted in fracture to the hip and none produced discernable injuries to the knee or thigh. Specifically, three tests resulted in fractures of the femoral neck and the other fourteen tests produced acetabular fractures.

The average age of the sixteen subjects used in the seventeen tests with usable hip tolerance data is 68 years (sd = 13 years). The average bone condition is in the normal range (*t-score* ≥ -1.0), although some of the female specimens showed evidence of osteopenia or minor bone degeneration (-2.0 ≤ *t-score* < -1.0).

The average tolerance of the hip from the seventeen tests that produced hip fracture is 5.70 kN (sd = 1.38 kN). In computing this average, data from pairs of left and right KTH tests from the same subject (5L and 5R, 8L and 8R, 25L and 25R) were averaged prior to calculating the mean hip tolerance. This ensured that the tolerances from these subjects were not over weighted with respect to the total subject population.

Figure 14 shows force histories from all nineteen of the tests, and indicates that most of the times-to-peak force (i.e., times-to-fracture force, except in the case of test 13R where no fracture occurred) from these tests are within the desired range of 20-60 ms, and that the loading rates are generally below the maximum desired loading rate of 300 N/ms, as indicated by the dashed line. Loading rate data in Figure 14 are divided into a higher loading rate group consisting of six curves, and a lower loading rate group consisting of the remaining 13 curves. Data from tests of truncated femurs (tests 5R and 5L, and 6R) fall within both the higher and lower loading-rate groups. Data from male and female subjects also fall within both groups. A further discussion of these two groupings is provided later in the paper.

TABLE 2. Results of Hip Tolerance Testing

Test ID	Force at Fracture (kN)	Time To Peak (ms)	Loading Rate (N/ms)	Calculated KTH Stiffness (N/mm)	Fractures
5L	5.59	13.7	361	NA*	Acetabulum ("T-type fracture"), inferior ramus
5R	5.37	15.8	303	NA*	Acetabulum (transverse, posterior wall)
6L	4.85	33.3	175	NA*	Acetabulum (posterior wall)
7R	4.49	38.6	114	208	No Injury
8L	7.52	33.9	417	334	Femoral neck
8R	7.87	23.5	566	534	Femoral neck
10L	6.60	29.5	326	379	Acetabulum (posterior wall), pubic rami
12R	6.67	56.1	138	195	Acetabulum (posterior col., anterior hemi-transverse fx.) pubic rami
13R	3.34	53	93	105	Iliac wing, pubic rami
14R	4.65	36.1	146	191	Femoral neck
16R	5.59	45.7	125	197	Acetabulum (posterior wall/column), inferior pubic ramus
17L	4.79	38.5	80	119	Acetabulum (transverse posterior wall), inferior pubic ramus
18L	5.57	40.4	159	249	Acetabulum (posterior wall)
19R	4.04	31.3	161	NA*	Acetabulum (posterior wall/column)
22L	8.85	33.6	326	268	Acetabulum (posterior rim)
24R	3.91	34.5	144	181	Acetabulum (transverse posterior wall), Pubic rami
25L	5.67	55.3	132	189	Acetabulum ("T-type" with comminuted posterior wall)
25R	5.87	59.2	132	177	Acetabulum (posterior rim)
26L	6.60	54.7	138	172	Acetabulum (posterior wall, anterior/superior rim)
Mean	*5.70*†	*38.3*†	*193*††	*233*†††	
sd	*1.38*†	*11.5*†	*114*††	*110*†††	

*NA=Not applicable because of invalid ram displacement measurements or because a whole KTH was not tested.
† Calculated using averages of data from subjects where both left and right sides were tested and excluding tests 7R and 13R where no hip fractures occurred.
†† Calculated using averages of data from subjects where both left and right sides were tested.
††† Calculated using averages of data from subjects where both left and right sides were tested and excluding all tests where stiffness could not be calculated due to missing ram displacement measurements.

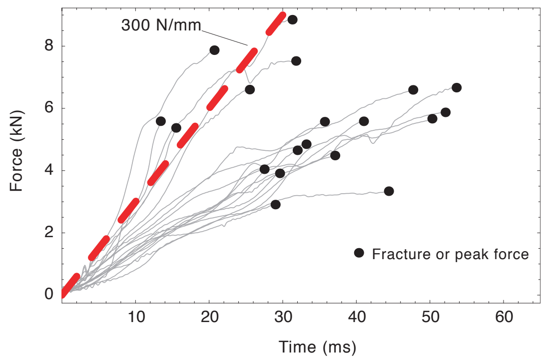

FIGURE 14. Applied force histories from hip tolerance tests, illustrating observed loading rates and times-to-peak force.

As shown in Figure 15, the distribution of different hip and pelvis injuries from these hip tolerance tests closely matches the distribution of hip/pelvis injuries in frontal crashes from the UM CIREN database. Multiple hip/pelvic injuries occurred in 35% of the CIREN cases compared to 23% of the hip tolerance tests. Table 3 illustrates that the percentages of the specific combinations of different pelvic injuries produced in these tests are also similar to those in UM CIREN cases. Sacroiliac injuries are excluded from the comparisons in Figure 15 and Table 3 because these injuries cannot occur in the hip tolerance tests because the iliac wings are rigidly fixed. In both the hip tolerance and UM CIREN datasets, the majority of the fractures are to the posterior wall, rim, and/or column of the acetabulum. This comparison suggests that the methods used to grip the pelvis resulted in realistic boundary conditions near the acetabulum and reasonable loads throughout the rest of the pelvis.

The specific nature of the fractures produced in hip tolerance testing is also similar to injuries observed in the CIREN data. For example, Figure 16 shows a CT scan of a typical acetabular rim fracture in the CIREN database produced by knee loading sustained in an offset-frontal crash, in comparison to an acetabular rim fracture from the hip tolerance testing. Both the location of the injury (i.e., at the posterior rim) and the size of the fragment separated from the acetabulum are similar.

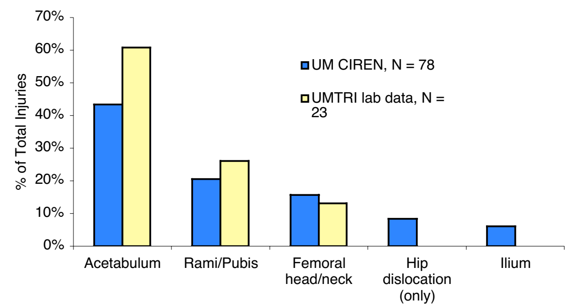

FIGURE 15. Distributions of hip and pelvis injuries from hip tolerance tests and frontal crashes in the UM CIREN database

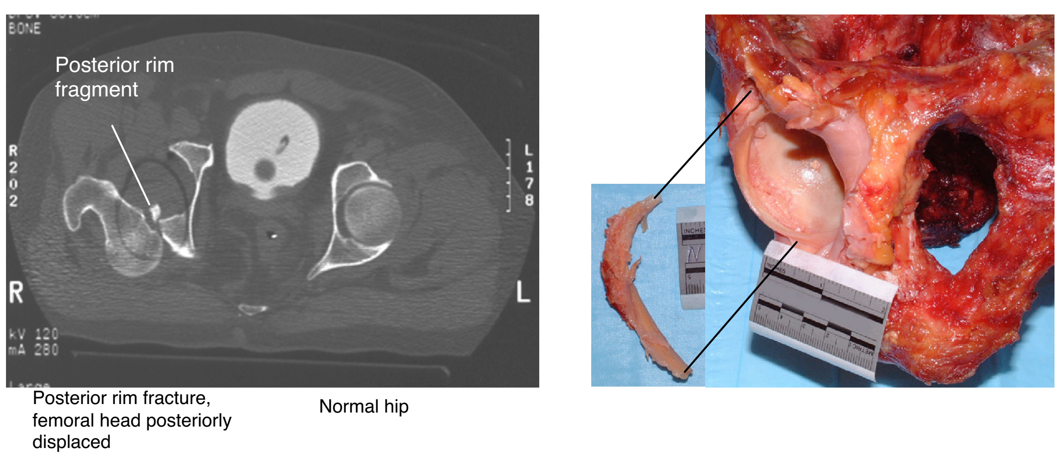

FIGURE 16. CT scan showing cross section of posterior-rim fracture from an offset frontal crash in the UM CIREN database (left) and photograph of posterior-rim fracture from a hip tolerance test (right). Fracture location and the size of the piece of acetabulum broken off are similar.

TABLE 3. Percentages of Different Combinations of Hip/Pelvis Fractures in UM CIREN Frontal Crashes and Hip Tolerance Tests

Injury Combinations	UM CIREN	Test Data
Acetabular fracture with or without dislocation	38%	47%
Pubic rami + acetabulum fractures	25%	35%
Femoral head/neck fractures	13%	18%
Femoral head/neck with acetabular fracture	13%	0%
Hip dislocation	5%	0%
Other	6%	0%

Effect of a Contralateral Injury on Tolerance and Injury Patterns

Some of the fracture forces listed in Table 2 were obtained from tests performed on specimens with preexisting injuries to the contralateral hip that were produced in other hip tolerance tests at different postures. Because a preexisting injury to the contralateral hip has the potential to weaken the whole pelvis and thereby affect tolerance of the other uninjured hip, neutral posture tests were conducted on both the left and right KTH complexes of three cadavers to explore this effect. The fracture tolerances from these tests are listed in Table 4 and the injuries produced are listed in Table 5. The average difference in fracture tolerance between first and second tests is approximately 4%, and injuries produced in the first and second tests (i.e., the left and right sides) of the same subject for two of the three subjects are comparable. These results indicate that hip tolerance in the second test on a specimen is not substantially affected by a hip fracture produced in the previous test of the opposite side.

TABLE 4. Fracture Tolerance for First and Second Tests on a Single Subject

Test ID	First Test	Second Test	Percent Difference
5L, 5R	5.59	5.37	4.00%
8L, 8R	7.52	7.87	-4.45%
25L, 25R	5.67	5.87	-3.41%

TABLE 5. Fractures Produced in Left and Right Hip Tolerance Tests on the Same Subject

Test ID	First Test Injuries	Second Test Injuries
5L, 5R	"T-type" acetabular	Transverse acetabular, posterior wall
8L, 8R	Femoral neck	Femoral neck
25L, 25R	"T-type" fracture with comminuted posterior wall	Posterior rim

Inertial Effects

As expected, inertial effects, which would manifest as differences between the applied and reaction force histories, are very small in these fixed-pelvis hip tolerance tests. For example, Figure 17 shows a typical set of force histories from the inertially compensated ram load cell, which measures applied force, and the load cell behind the pelvis, which measures reaction force. As indicated, these curves are nearly identical, with the only difference being that the reaction force lags the applied force by several milliseconds. This lag was hypothesized to occur based on data from previous biomechanical studies and is thought to result from the compliance of the knee and hip joints.

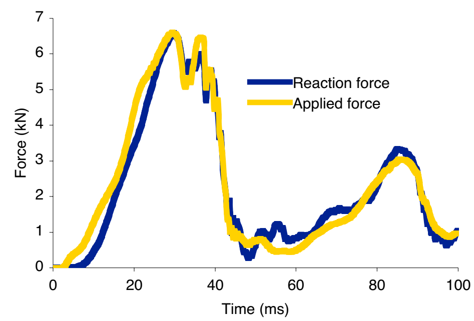

FIGURE 17. Applied and reaction force histories from a typical hip tolerance test.

Effects of Subject Characteristics on Hip Tolerance

The fracture forces in Table 2 were analyzed with specimen data to determine if age, gender, stature, body mass, and *t*-score are associated with hip tolerance. For these analyses, data from pairs of left and right tests on a single subject were averaged and used as a single data point, and data from tests 7R and 13R were again excluded, resulting in data from four female and ten male subjects for the analysis. As indicated below, no significant associations were found.

Figure 18 shows the mean hip tolerance for male cadavers and female cadavers. The average hip fracture tolerance for the male cadavers is 5.96 kN (sd = 0.61 kN) and the average fracture tolerance for the female cadavers is 5.37 kN (sd = 0.49 kN). While the average tolerance for female specimens is slightly lower than that for male specimens, the difference is not statistically significantly ($p = 0.49$).

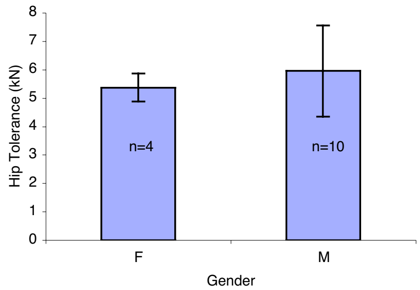

FIGURE 18. Hip tolerance by gender. Male tolerance (n = 10) was not found to be significantly different than female tolerance (n = 4, $p = 0.49$).

Figures 19 and 20 show the distribution of hip fracture tolerance by stature and body mass. There is little association between these parameters and hip tolerance ($R^2 = 0.04$ for stature, and $R^2 = 0.07$ for mass). As a result, mass- or stature-based scaling techniques have not been used to normalize the hip tolerance data in these tests.

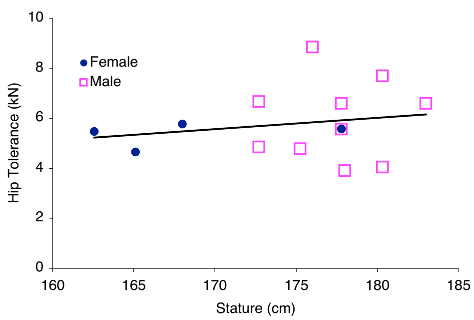

FIGURE 19. Hip tolerance by stature and gender ($R^2 = 0.04$).

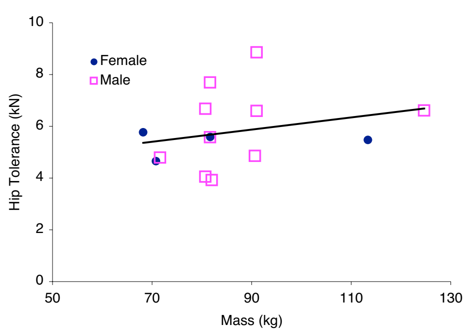

FIGURE 20. Hip tolerance by mass and gender ($R^2 = 0.07$).

Figure 21 illustrates the relationship between hip fracture tolerance and *t-score* using data from two female and nine male subjects for which osteogram and *t*-score data are available. As with the other subject variables, there is little association between *t-score* and hip fracture tolerance ($R^2 = 0.15$). In addition, no association was found between age and hip fracture tolerance ($R^2 = 0.13$).

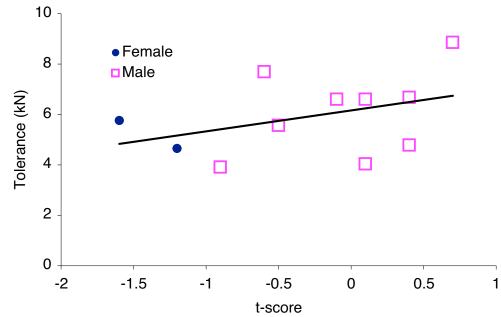

FIGURE 21. Hip tolerance by t-score and gender ($R^2 = 0.15$).

KTH Stiffness

Using the displacement of the ram and the force applied to the knee, the stiffness of the KTH complex with a fixed pelvis boundary condition was calculated. The average stiffness of the KTH complex is 233 N/mm (sd = 110 N/mm). For reference, the stiffness of the test fixture, measured using a Hybrid III midsize male pelvis gripped at the iliac wings is 2000 N/mm.

Femur Tolerance and Injury Patterns

The average age of the sixteen subjects used in the hip tolerance tests is 68 years (sd = 15 years). The average bone condition is in the normal range (*t-score* ≥ -1.0), although one of the female specimens showed evidence of osteopenia.

Table 6 lists the results of the thirteen femur tolerance tests conducted on specimens from seven cadavers, along with the hip fracture forces from tests on the same subjects. The hip fracture tolerance listed for tests 25RF and 25LF is the average of the hip tolerances from tests 25R and 25L.

As indicated, all of these tests produced femoral neck fractures, and two tests (on the left and right femur from a single subject) also produced distal femoral shaft fractures. The character of the femoral neck fractures produced in the femur tolerance testing is similar to the femoral neck fractures produced in hip tolerance tests 8L and 8R. Figure 22 shows a typical fracture produced in the knee/femur tests relative to the femoral neck fracture produced in hip tolerance test 8R. The similarity between these two fractures indicates that the “acetabular cup” used in these tests provides boundary conditions that are representative of a fixed “acetabular” surface.

TABLE 6. Results from Femur Tolerance Testing

Test ID	Force @ Femur Fx. (kN)	Force @ Hip Fx. (kN)	Hip Tolerance as a %age of Femur Tolerance	Time To Peak (ms)	Loading Rate (N/ms)	Stiffness (N/mm)	Major Fractures
18RF	8.3	5.6	67%	35.8	212	N/A	Femoral neck
19LF	6.3	4.0	64%	35.6	153	330	Femoral neck
19RF	6.5	4.0	62%	36.8	174	420	Femoral neck
22LF	10.0	8.8	88%	31.8	420	594	Femoral neck
22RF	11.5	8.8	76%	20.9	789	505	Femoral neck
23LF	6.3	NONE	N/A	36.5	251	N/A	Femoral neck
23RF	6.9	NONE	N/A	45.6	159	N/A	Femoral neck
24LF	6.5	3.9	60%	36	234	249	Femoral neck
24RF	5.7	3.9	68%	36.4	191	214	Femoral neck
25LF	7.8	5.8*	74%	30.2	352	329	Femoral neck and distal shaft
25RF	7.4	5.8*	78%	27.6	396	344	Femoral neck and distal shaft
26LF	6.9	6.6	95%	28	376	340	Femoral neck
26RF	7.9	6.6	84%	26.9	473	315	Femoral neck
Mean	*7.59*†	*5.79*†	*76%*†	*33.1*†	*314*†	*364*††	
sd	*1.58*†	*1.82*†	*11%*†	*5.6*†	*161*†	*116*††	

*Obtained by averaging the results of tests 25L and 25R.
†Calculated using averages of data from subjects where both left and right sides were tested.
††Calculated using averages of data from subjects where both left and right sides were tested and excluding all tests where stiffness could not be calculated due to incorrect deflection measurement.

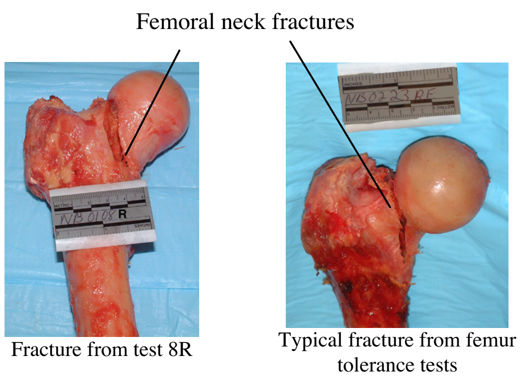

FIGURE 22. Fracture pattern produced in hip tolerance test 8R and typical fracture from femur tolerance tests.

The fracture forces for the left and right sides of the same subjects are generally comparable, with the largest difference being 1.5 kN between tests 22RF and 22LF. A single tolerance value was determined for each subject by averaging results of left- and right-side femur tolerance tests. This ensured that each subject was given equal weight and resulted in an overall average femur tolerance force of 7.59 kN (sd = 1.58 kN). Figure 23 shows the acetabular fracture force, listed in Table 6, as a percentage of the femoral neck failure force for each subject. As indicated, the acetabular fracture force is always less than femur fracture force, and, on average, acetabular fracture load is 76% (sd = 11%) of the femoral neck failure load. A paired t-test shows that acetabular failure force is significantly different than femoral neck failure force ($p < 0.05$).

Figure 24 is a plot of acetabular versus femoral neck fracture force for the six test subjects for which hip (i.e., acetabulum) and femur tolerances are available. As indicated, there is a strong correlation between the two tolerance measures ($R^2 = 0.88$).

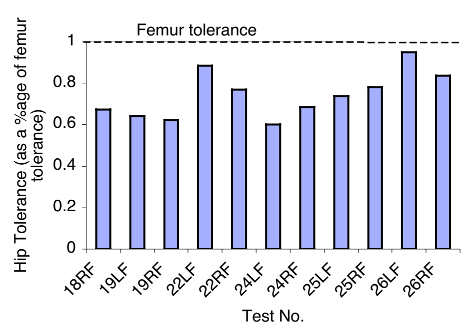

FIGURE 23. Acetabular tolerance relative to femoral neck tolerance.

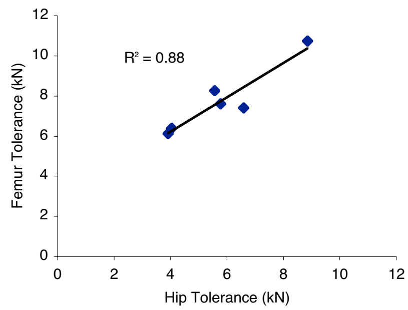

FIGURE 24. Acetabular tolerance versus femoral neck tolerance for the six test subjects where acetabular tolerance was available.

Femur Stiffness

Using the displacement of the ram and the force applied to the knee, the stiffness of the knee/femur complex was calculated for each femur tolerance test. The average stiffness for all subjects is 364 N/mm (sd = 116 N/mm). As expected, this is significantly higher than the average KTH stiffness of 233 N/mm (sd = 110) determined in hip tolerance tests.

DISCUSSION

The NASS database was analyzed to explore the relative incidence of knee, thigh, and hip injuries in real-world frontal crashes and crash, restraint, and occupant characteristics that are associated with these injuries. This analysis indicates that hip injuries are occurring to occupants of all ages restrained by both three-point belts and airbags, and in frontal crashes that are similar or of lesser severity than FMVSS 208 or NCAP testing.

It is hypothesized that previous biomechanical research involving frontal-impact loading of cadaver knees failed to produce hip fractures because high loading rates caused fractures of the knee or femur before enough time (e.g., 2-5 ms) had elapsed to allow transmission of force to the hip. The short times to knee and distal femur fracture in these earlier studies result from relatively high loading rates that are not representative of knee loading rates in airbag/knee bolster equipped vehicles. Because these earlier tests produced knee and femur fractures, it was assumed that the hip has a higher tolerance than the femur or knee, when it actually has a lower tolerance as demonstrated by the results of the current study.

As the first step in developing an improved injury prediction model for the KTH complex in frontal crashes, the dynamic fracture tolerance of the hip to loading applied at the knee and directed toward the hip has been determined using data from tests performed on sixteen unembalmed human cadavers. For these tests, the orientation of the pelvis relative to the femur was set to represent that of a normal driving posture. Tests were conducted with the pelvis fixed to minimize inertial effects and thereby allow the weakest portion of the KTH complex to fail first. In addition, the fracture tolerance of the femur has been determined by testing the knee/femur complexes from the subjects used in hip tolerance testing under similar loading conditions.

Loading rates for all hip tolerance tests were generally within the target range (below 300 N/ms). However, the distribution of loading rates shown in Figure 14 is stratified into two groups, one of which is close to the upper bound of the target range. These groupings are not due to differences in test conditions, since all tests were conducted using similar loading conditions, and the three tests with truncated femurs are distributed between the two groups. Also, these groupings do not appear to be from differences in subject gender, age, stature, and mass, since none of these show a significant association with loading rate (R^2 values for all of these associations are less than 0.1). It is believed that differences in loading rates are from inter-subject differences in KTH axial stiffness. This is supported by the strong association between loading rate and KTH stiffness ($R^2 = 0.88$). However, the reason for the distinct inter-subject differences in stiffness is unknown.

The results of the hip and femur tolerance tests demonstrate that, under the applied loading conditions, the hip is weaker than the knee or thigh, and that the weakest portion of the hip is the posterior acetabulum. For the cadaveric specimens used in this study, the failure tolerance of the acetabulum is 5.70 kN (sd = 1.38 kN). By comparison, the weakest portion of the femur under these loading conditions is the neck, which has a fracture tolerance of 7.59 kN (sd = 1.58 kN). These results imply that, under distributed knee loading, the fracture tolerance of the more distal portions of the femur and the knee (i.e., the midshaft, supracondylar, and condylar regions of the femur, and patella), upon which the current FMVSS 208 KTH injury criterion of 10 kN is based, have a higher fracture tolerance than either the femoral neck or acetabular portion of the hip.

If the tolerance of the hip is significantly lower than the tolerance the of thigh or knee, one might expect the occurrence and risk of hip injury in frontal crashes to be even more frequent and higher relative to the occurrence and risk of knee or thigh injuries than is found in the NASS database, as shown in Figures 2-6. The reason that hip injury occurrence and risk are not even higher relative to thigh and knee injuries is not yet fully understood, but answering this question is the focus of much of the remaining research in the UMTRI KTH research program. At this time, it is hypothesized that the axial compliance of the KTH complex and the relative effective masses of the leg, KTH, and torso are potentially important factors that affect the relative magnitudes of forces along the KTH from knee loading and the time lags in these forces. Also, effective masses are expected to depend on body posture, muscle tension, and loading rates. In addition, the importance of time lags relative to forces exceeding the various KTH tolerance levels may also depend on the rate at which the force is applied at the knee. Hip posture is also a factor that has the potential to increase the incidence of hip injury, relative to knee or femur injury, by decreasing the tolerance of the hip.

The majority of the fractures produced in the femur tolerance testing were to the femoral neck. However, in the field, other more distal parts of the femur are often injured. The reasons that femoral neck fractures do not occur more frequently in the real world are thought to be due to inertial effects in the thigh and the orientation of the femur relative to the direction of load. Inertial effects decrease the force that is transmitted from the knee, along the femur, to the hip such that the tolerance of the more distal femur is exceeded prior to the tolerance of the femoral neck being exceeded. Changes in the orientation of the femur relative to the direction of loading could also change the part of the femur that is fractured by increasing the tendency of the femoral shaft to fail in bending.

Because of the lack of hip tolerance data for frontal knee loading in the literature, it is not possible to compare the results from the current study with existing data. There are, however, femur tolerance data available for frontal impact loading, although the loading conditions are somewhat different from those used in the current study. For example, Carothers et al. (1949) performed quasi-static testing on embalmed femurs supported in an acetabular-like cup and found the femur tolerance to be 8.6 ± 1.8 kN. This tolerance is not significantly different from that of the current study ($p = 0.24$). In addition, four out of the five tests conducted by Carothers et al. resulted in femoral neck fractures that are similar to the femoral neck fractures produced in the testing reported on in this paper.

Subsequently, Viano and Stalnaker (1980) performed rigid and padded, flat-faced high-speed (about 13 m/s) pendulum impacts to the knees of six human cadavers with denuded femurs and suspended in a free-back condition. Denuding the femur greatly reduced the mass of the KTH complex and consequently reduced inertial effects on force reduction along the KTH complex. When the results from two osteoporotic subjects are removed from these data, the tolerance of the femur is 7.7 ± 1.8 kN, which is not statistically different from the current femur tolerance ($p = 0.88$).

It is possible that the fracture forces measured during femur tolerance tests in the current study may have been reduced by the previous sub-fracture femur loading applied in the hip tolerance tests that were conducted on the same specimens. However, the similarities between the fracture tolerances in the current dataset and the comparable Viano and Stalnaker (1980) data suggest that this is not the case.

While the femur tolerance data in the current study are comparable to those measured in previous studies, the elderly subject populations used in the current and previous studies are expected to have a lower fracture tolerance than the average adult in the driving population, which is almost thirty years younger than the average age of the subject population of the current study. Because of this, the tolerances presented in this paper should be considered conservative estimates for the total adult occupant population.

The reasons for a lack of association between subject stature, age, body mass, gender, or *t-score* and hip tolerance are unclear, but may result, in part, from the relatively small sample sizes and/or small variances in these subject variables. The relatively strong association between hip (i.e., acetabular) and femur tolerance across subjects suggests that these tolerances are subject specific. A better association between hip tolerance and one or more of these subject variables would therefore be expected. It should also be noted that the frequency of hip (i.e., primarily acetabular) fractures in frontal crashes in the UM CIREN database are not associated with occupant gender or age. It is therefore possible that the nature of these types of hip injuries is such that it is a function of subject- or crash-specific factors that have not yet been identified.

The femur stiffness data provided by this study should be useful for development and validation of ATDs and mathematical models of the KTH. The average stiffness of the knee/femur complex was found to be 364 N/mm (sd = 116 N/mm). Because the molded knee interface distributed the applied loads over the patella and femoral condyles, the knee compliance is not expected to substantially contribute to the measured knee/femur stiffness. Therefore, because of the bowed shape of the femur and the direction of applied load along the long axis of the femur, the measured stiffness probably represents a combination of the bending and compressive stiffness of the femur alone and should only be applied to the design of this segment of ATD and FEM models.

Future research will explore inertial and compliance effects in the KTH using both modeling and experimental techniques. In addition, the effects of hip flexion and thigh adduction from the neutral posture, which are hypothesized to decrease the tolerance of the hip by reducing the contact area between the femoral head and acetabular surface, are currently being investigated. The results of this ongoing research will be reported in a future paper, along with a proposal for a new KTH injury criterion for frontal crash protection.

CONCLUSIONS

As a first step in a program to develop a new KTH injury prediction model, an experimental method has been developed for determining the lowest fracture tolerance of the KTH complex for impact loads applied to the knee and directed toward the hip joint center. Loading rates used are similar to those that occur during knee-bolster loading in frontal crashes of late-model vehicles. In contrast to the previous belief that the hip is stronger than the femur, the results of tests conducted using KTH specimens from unembalmed cadavers demonstrate that the acetabular region of the hip fails at a lower force than the more distal portions of the KTH complex. In addition, tests conducted with the acetabulum and pelvis replaced by a fixed "acetabular cup" demonstrate that the femoral neck is the weakest part of the femur under distributed knee loading (i.e., the tolerance of the femoral shaft and supracondylar region are higher than the tolerance of the femoral neck). For the tests conducted in this study, the fracture tolerance of the hip is 5.70 kN (sd = 1.38 kN), while the fracture tolerance of the femur is 7.59 kN (sd = 1.58 kN).

The fixed pelvis test methods used in this study to determine hip tolerance were validated by comparing the specific types of hip injuries produced by the current testing to hip injuries from real-world frontal crashes documented in the UM CIREN database. Both the distribution of hip/pelvis fractures and the specific nature of the acetabular fractures are similar for the test data and the crash-injury data. In both datasets, the most common hip injury is the fracture of the posterior wall of the acetabulum.

While the data show a good correlation between the fracture tolerance of the posterior acetabulum and the fracture tolerance of the femoral neck across test subjects, no associations between subject factors, including age, bone condition, stature, total body mass, or gender, and fracture tolerance of the hip were found in these data. The reasons for the lack of association with these subject factors are not clear, but the occurrence of real-world hip injuries in the NASS database are also not specific to these particular subject factors.

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