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*INJURY BIOMECHANICS RESEARCH*
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# Development of an Experimental Protocol to Quantify the Tolerance of the Knee-Thigh-Hip Complex to Knee Loading in Frontal Impacts

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S. Kuppa, and L. W. Schneider

## ABSTRACT

*Injuries to the knee-thigh-hip (KTH) complex from loading to the patella in frontal and offset-frontal automotive crashes are a significant problem with large societal costs. Previous biomechanical research has focused on crash-induced injuries to the knee and femur rather than injuries to the hip because knee and femur injuries have historically been more common and because of difficulties producing hip injuries from knee loading in the laboratory. Government safety standards and improvements in motor-vehicle design over the last two decades, particularly the introduction of energy-absorbing knee bolsters, have decreased the relative incidence of knee and femur injury in frontal crashes. However, the relative incidence of hip injury has increased. This trend is of particular concern because hip injury frequently results in long-term impaired mobility and has a higher associated rate of mortality than other KTH injuries.*

*The goal of this project is to develop a new comprehensive criterion that describes the potential for KTH injury from knee loading in a frontal motor-vehicle crash. Because previous research has documented the tolerance of the knee and femur to this type of loading, the proposed research primarily focuses on determining the fracture/dislocation tolerance of the isolated hip and the hip as a part of the KTH complex.*

## INTRODUCTION

Based on an analysis of data in the National Automotive Sampling System database from 1995-2000, approximately 30,000 occupants sustain AIS 2+ (AAAM 1990) fractures and dislocations to the knee-thigh-hip (KTH) complex annually in frontal crashes. Of these injuries, approximately 14,000 are to the hip complex, which includes the femoral neck, femoral head, and pelvis. Most hip injuries are clinically more severe and more difficult to treat than injuries to either the knee or thigh, and can result in lifelong impaired gait (Nerubay et al. 1973). In addition, hip injuries account for the majority of life years lost to KTH injury in motor-vehicle crashes (Kuppa et al. 2001). The mortality rate from hip injuries is estimated to be as high as 24% for persons over age 50 and the total annual cost of hip injuries (excluding costs associated with rehabilitation and lost wages) has been estimated at over $400 million (Office of Technology Assessment 1994).

Early studies of the tolerance of the KTH complex to impact at the patellar surface of the flexed cadaver knee produced mostly patellar and distal femur fractures and few hip fractures (Patrick et al. 1966; Powell et al. 1974 and 1975; Melvin et al. 1975; Melvin and Stalnaker 1976). The low incidence of hip fractures in these early studies was thought to indicate that the tolerance of the hip to knee impact is greater than that of the femur or knee. This hypothesis was supported by accident data from the same time period (pre 1975), which showed that femur fracture occurred in 60% of crashes that resulted in KTH injuries (Melvin and Stalnaker 1976). Because the femur was thought to be the weakest link in the KTH complex, the dynamic tolerance of the femur was thought to be an injury criterion capable of protecting the entire KTH complex. Consequently, femur tolerance data collected from the dynamic knee loading performed by Patrick, Powell, and Melvin were used to select a maximum force 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 the midshaft femur of a midsize-male Hybrid III anthropomorphic test device (ATD), or crash test dummy, must not exceed 10 kN in either a 30-mph full-frontal soft-pulse sled test with an unrestrained dummy or in a restrained 30-mph full-vehicle barrier impact with a belt-restrained dummy. The 10-kN tolerance was later correlated to a 35% risk of injury to the KTH complex (Morgan et al. 1989).

Figure 1 shows the results of a recent analysis of the University of Michigan (U of M) Crash Injury Research and Engineering Network (CIREN) database, which suggest that the incidence of hip fracture in frontal and offset-frontal crashes is higher in newer model vehicles, relative to knee and femur fractures, than in earlier models. A further analysis of the hip injuries in this dataset shows that hip injuries are occurring in frontal crashes that are less severe than regulatory compliance tests specified in FMVSS 208 (i.e., less than 30-35 mph deltaV). A possible reason for this is shown in Figure 2, which indicates that AIS$\geq$2 hip injuries tend to occur on the side of the body toward which the injured occupant tends to move. That is, occupants who tended to move forward and to the right in a crash sustain mainly right hip injuries, and occupants who tended to move forward and to the left sustain left hip injuries.

The association between the side of hip injury and the direction of occupant motion is hypothesized to result, in part, from femur adduction caused by occupant motion, which tends to decrease the contact area of the femoral head on the acetabular surface and consequently increase the localized stress on the acetabulum (Letournel and Judet 1993). It is also possible that hip fractures are from the higher load experienced by the KTH on the side of the body corresponding to the direction of occupant motion (e.g., an occupant moving forward and to the right would preferentially load the right knee). In addition, it is believed that hip flexion caused by the occupant ramping up the seat pan and by forward rotation of the occupant's torso can result in decreased contact area between femoral head and the acetabular surface, which consequently, can decrease the injury tolerance of the hip. An analysis of all available cadaver knee impact data (Patrick et al. 1966; Powell et al. 1975; Melvin and Stalnaker 1976; Melvin and Nusholtz 1980; Leung et al. 1983; Donnelly and Roberts 1987) supports the hypothesis that a change in thigh orientation in a frontal crash leads to hip fracture. The few hip fractures that occurred in these studies were generally produced when the orientation of the thigh was changed so that flexion or adduction occurred either during or prior to impact.

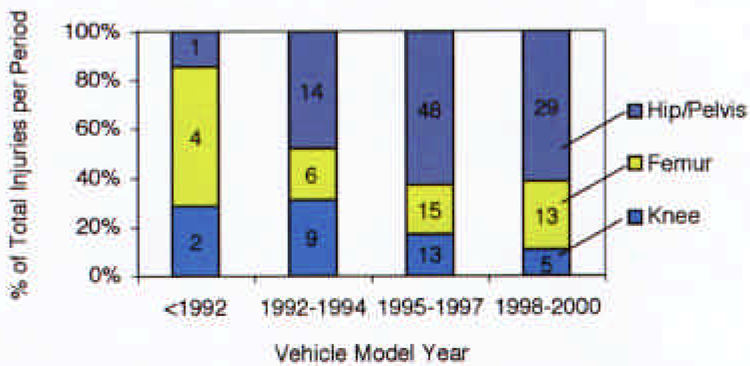


Figure 1. Relative incidence of AIS≥2 KTH injuries in U of M CIREN database (n=159).

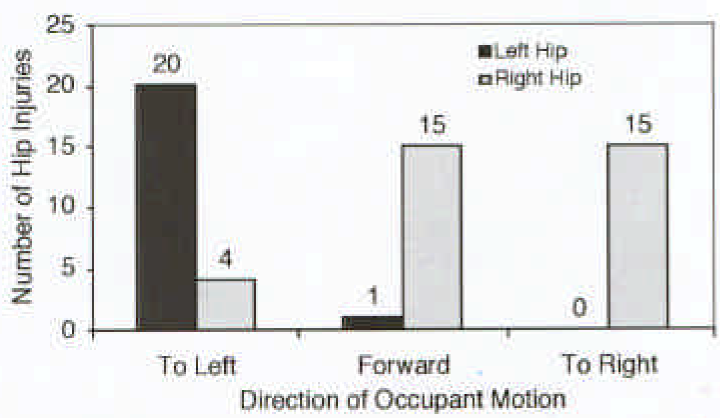


Figure 2. Distribution of AIS≥2 hip injuries by direction of occupant motion.

One of the major shortcomings of previous knee-impact studies is that the rates of force application were almost always higher, and the duration of the applied load shorter, than what is observed during frontal crashes involving late-model-year vehicles. The majority of the applied force histories in the studies used to develop the FMVSS 208 KTH injury criterion caused injury within 10 ms after the start of force application (Viano 1977), at loading rates varying between 400 and 3000 N/ms. Typical force histories for Hybrid III ATD loading current model knee bolsters tend to peak between 20 and 60 ms at loading rates below 300 N/ms. These differences are not large enough to result in substantial changes in bone tolerance due to viscoelastic effects (McElhaney 1966). However, it is hypothesized that they are large enough to cause differences in inertial effects that could alter the distribution of forces along the KTH complex. That is, higher loading rates generate a larger difference in forces at the knee relative to those at the midshaft femur, while lower loading rates generate less of a difference between forces at the knee and midshaft femur (Horsch and Patrick 1976; Donnelly and Roberts 1987). Using rigid, flat-faced pendulum impacts that are typical of the knee impacts used to develop FMVSS 208, a 10-kN load applied to the knee produced approximately 5.3 kN at the cadaver midshaft femur. This 47% reduction in force is due to the acceleration of the mass of the distal KTH complex, i.e., it is proportional to the mass between the knee and the midshaft femur (Donnelly and Roberts 1987).

Since there is mass between the midshaft femur and the hip, the forces generated at the hip by dynamic knee loading can be assumed to be less than the force at the midshaft femur. These differences are proportional to the acceleration and mass of the leg. If the rate of force application at the knee changes, then relationships between force levels at the knee, thigh, and hip will also change. In other words, a lower rate of loading could result in lower forces at the knee and midshaft femur, and relatively higher forces at the hip due to a reduction in inertial effects at lower loading rates. This would be inconsequential if the tolerance of the hip is higher than that of the femur or knee, as the femur or knee would always fail first. However, the incidence of hip injury in real-world crashes suggests that this is likely not the case, i.e., the tolerance of the hip is probably less than that of the knee or femur. Consequently, the current FMVSS 208 KTH injury criterion, which is based on femur fracture at high loading rates, may not adequately protect the entire KTH complex under the lower loading rates typically generated in crashes involving newer model vehicles with energy-absorbing knee bolsters.

The proposed research is designed to generate data on hip injury tolerance as a function of thigh orientation. These data will be analyzed and combined with previously reported injury tolerance data for the femur and knee, and with relationships between knee loading rate and force distribution along the KTH. The result will be a new KTH injury-prediction model that allows prediction of the likelihood of knee, thigh, and hip injuries in frontal crashes as a function of the rate and magnitude of applied force, as well as the orientation of the femur and the direction of the applied force. It is expected that this more comprehensive injury-prediction model for the KTH complex will allow the development of improved knee-bolster designs and other countermeasures that will reduce the likelihood of debilitating hip fractures and dislocations in frontal crashes.

## METHODS

Figure 3 shows a flow chart that outlines the steps involved in developing an improved KTH injury prediction model. Analyses of the CIREN and NASS databases and a review of the biomechanical literature were used to document the frequency of KTH injury and to provide insight on the mechanisms of injury. Hip tolerance tests using isolated cadaver KTH sections are being conducted to determine the tolerance of the hip as a function of femur orientation independent of the knee and thigh. The tolerance of the KTH complex determined from these tests will be compared to existing tolerances of the knee and femur from the biomechanical literature. The results from experimental testing, knee and femur tolerance values from the biomechanical literature, and the results of a study to explore loading-rate effects will be combined to develop a new KTH injury-prediction model and failure criteria for knee, femur, and hip fractures.

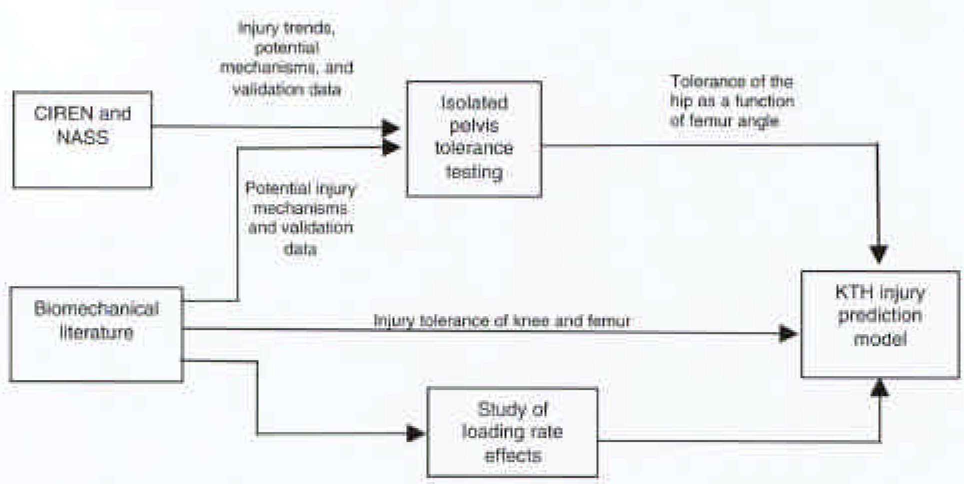


Figure 3. Flow chart detailing the steps to develop a new KTH injury criterion.

To date, preliminary analyses of the CIREN and NASS databases and a review of the biomechanical literature have been completed and two independent factors that have the potential to influence injury to the KTH complex have been identified. The first is loading rate, which is hypothesized to affect the relationship between forces applied to the knee and forces observed at the midshaft femur and hip. The second is thigh orientation relative to the pelvis and direction of loading, which is thought to influence the hip injury tolerance by altering the contact area between the femoral head and the acetabular surface.

Tests of isolated unembalmed cadaver pelvises are being conducted to determine the tolerance of the hip as a function of femur orientation for dynamic loading along the long axis of the femur. The isolated pelvis test device is illustrated in Figure 4. Prior to these tests, the pelvis and legs are removed from an intact, unembalmed cadaver. The pelvis is fixed to the test device by gripping the iliac wings. Pelvic rotation is prevented both by the fixed iliac wings and by a support at the pubic symphysis. A controlled force history is applied to the pelvis by a pneumatically accelerated weighted platform that contacts a linearly translating ram that is initially in contact with the knee (patella). In some cases, the leg and knee were removed by cutting through the mid shaft of the femur prior to testing with the isolated pelvis test fixture. When this was done, an interface fixture was molded to the mid shaft of the femur and the ram was positioned so that it initially contacted this fixture. The combination of a slow-moving (about 1 m/s) loading platform, energy-absorbing materials at the interface between the ram and the platform, and a fixed pelvis minimize inertial effects throughout the KTH complex. The combination of platform velocity, platform mass, and the character of the ram/platform interface have been selected to generate loading rates below 300 N/ms and times to peak failure force from 20 to 60 ms. These rates are more typical of loading rates and times to peak force measured at the Hybrid III femur in FMVSS 208 compliance testing with late-model vehicles. Applied force is measured at the ram/cadaver interface by a load cell attached to the ram. Reaction force is measured by a load cell positioned behind the pelvis. As illustrated in Figure 5, force is always applied along a vector connecting the midpoint of the lateral and medial femoral condyles and the hip joint center (i.e., generally along the long axis of the femur).

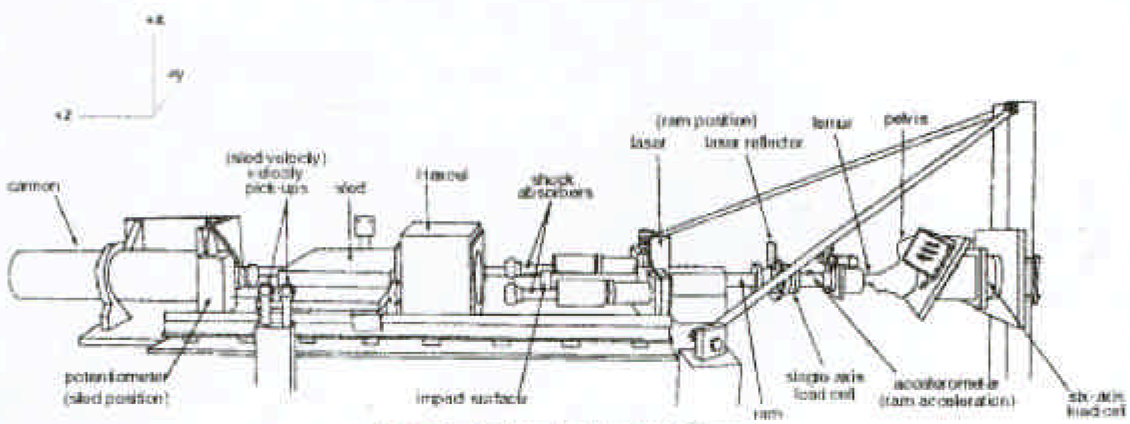


Figure 4. Isolated pelvis test fixture.

For the isolated pelvis tests, the zero-flexion condition is defined such that, in the side view, the angle between the long axis of the femur and the line formed by the plane defined by the ASIS points and the pubic symphysis is 120°. This corresponds to the amount of hip flexion in a standard automotive-seated posture defined by Schneider et al. (1983). The left side of Figure 6 shows a side view of the pelvis and femur orientations in this posture. For tests where additional hip flexion is simulated, the pelvis is either rotated prior to mounting, or aluminum wedges are inserted between the reaction force load cell and the hip-mounting fixture.

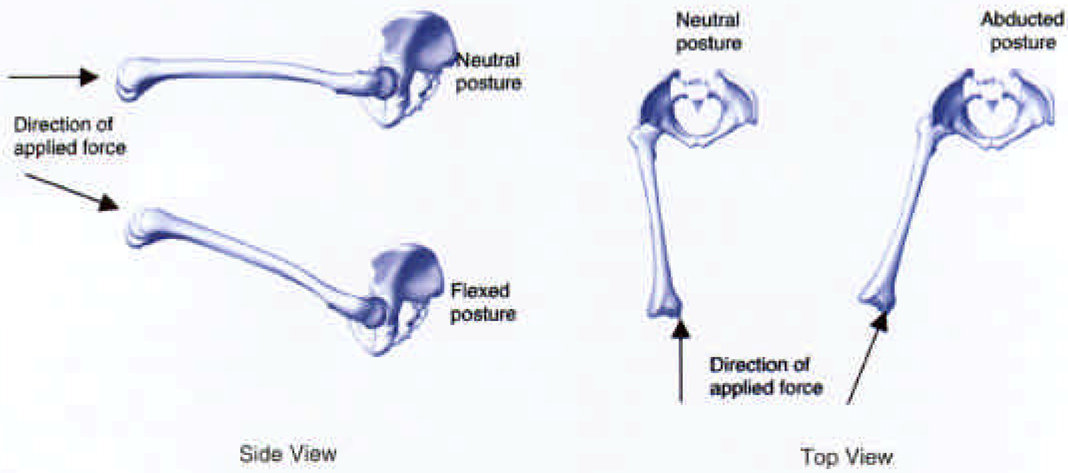


Figure 5. Variation in direction of applied force with posture. In the isolated pelvis test fixture the pelvis is rotated to achieve the flexion and adduction conditions shown above.

In a zero adduction test, the femur is oriented so that the axis defined by a line connecting the midpoint of the lateral and medial femoral condyles to the hip-joint center is perpendicular to the line connecting the left and right hip-joint centers. During a test, the hip-joint center location is estimated by palpating the head of the femur of the cadaver section in the isolated pelvis test fixture. The right side of Figure 6 shows a top view of the pelvis and femur illustrating this posture. For tests where adduction or abduction is simulated, aluminum wedges are inserted between the reaction force load cell and hip-mounting fixture to attain the desired abduction/adduction angle.

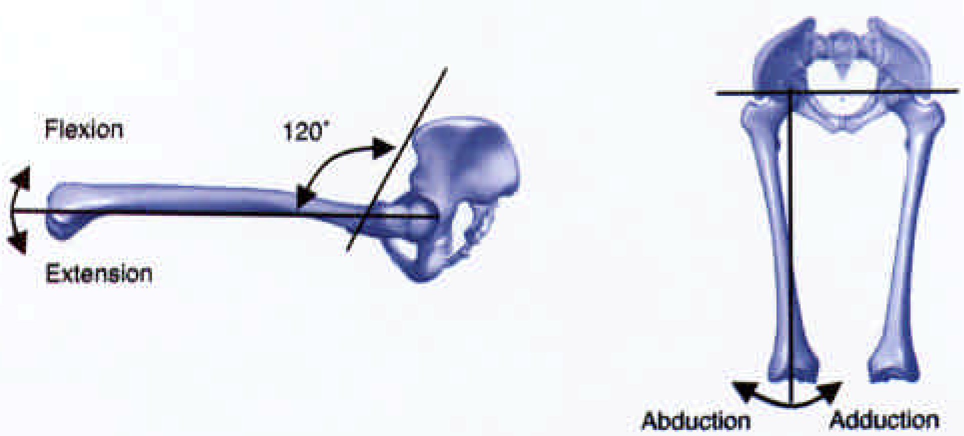


Figure 6. Definitions of hip flexion and abduction/adduction angles.

To maximize the potential for normal bone density, the age of male cadavers is limited to 85 years and the age of female cadavers is limited to 75 years, unless there is clear evidence of normal bone condition (e.g., a recent DEXA scan). In addition, subjects that died from diseases that could potentially affect bone condition are not used. To further ensure that data from osteoporotic subjects are not used, bone mineral density analysis in the form of osteograms (Yang et al. 1994) is conducted immediately following testing.

## RESULTS

To date, eight cadavers have been used in a total of 15 tests with the isolated pelvis test fixture shown in Figure 4. All of the tests produced injury to either the pelvis or hip. None of the tests produced discernable injuries to the knee or shaft of the femur. Importantly, as shown in Figure 7, the injury locations and patterns of hip/pelvis injuries produced in these tests closely resemble the distribution and character of hip/pelvis injuries in frontal crashes from the U of M CIREN database. In particular, the majority of the acetabular fractures produced in this dataset are to the posterior wall, rim, and column, which are the most frequently injured part of the pelvis in frontal crashes from the U of M CIREN database. This information confirms that methods used to mount the pelvis to the test fixture produce realistic boundary conditions near the acetabulum and a realistic load distribution throughout the rest of the pelvis. Injuries to the sacrum and pubic symphysis were not included in the analysis of the CIREN data shown in Figure 7 because these injuries are not likely to be produced in the current series of isolated pelvis tests. The character of the fractures produced in the testing is also similar to injuries observed in the CIREN data. For example, Figure 8 shows a CT scan of a typical acetabular rim fracture in the CIREN database produced by knee loading sustained in an offset-frontal crash relative to an acetabular-rim fracture produced in the isolated pelvis testing.

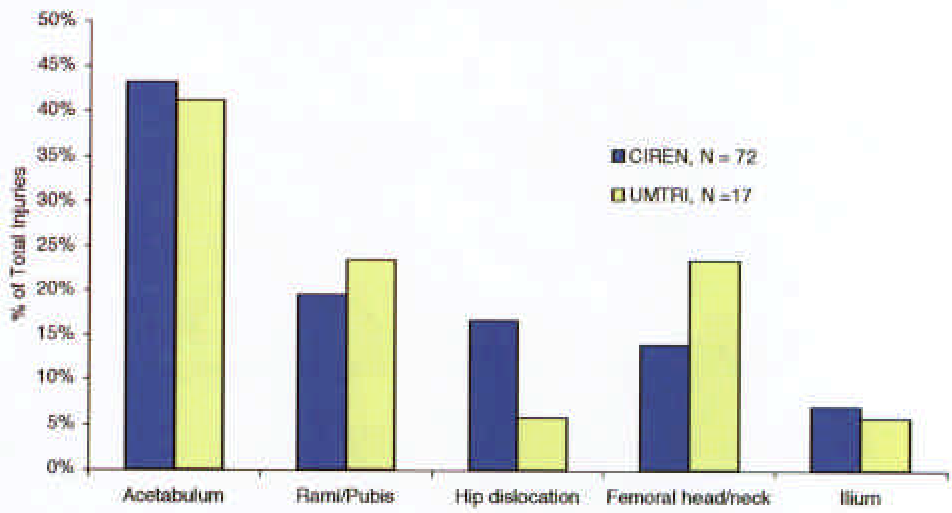


Figure 7. Comparison of the distributions of hip/pelvis injuries from the isolated pelvis tolerance tests and in the U of M CIREN database.

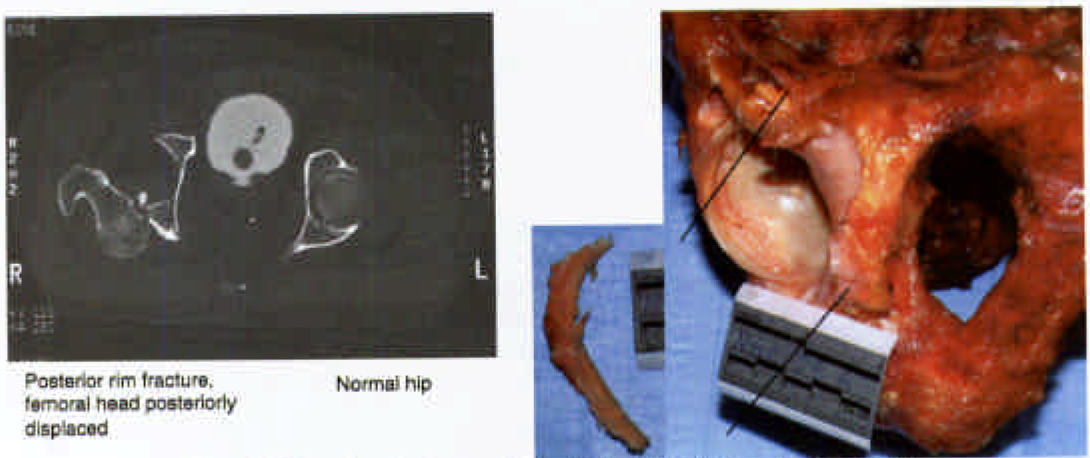


Figure 8. Posterior rim fracture in U of M CIREN (left) and posterior rim fracture in UMTRI test data (right).

Figure 9 shows a typical set of force histories from the load cell located on the ram, which measures applied force, and the load cell behind the pelvis, which measures reaction force. During the initial loading phase, the reaction force lags the applied force by several milliseconds. This is thought to be due primarily to laxity in the knee and hip joints. However, in general, the applied and reaction force curves match extremely well. This affirms the expectation that rigidly supporting the pelvis and applying a load at a low rate minimizes inertial effects.

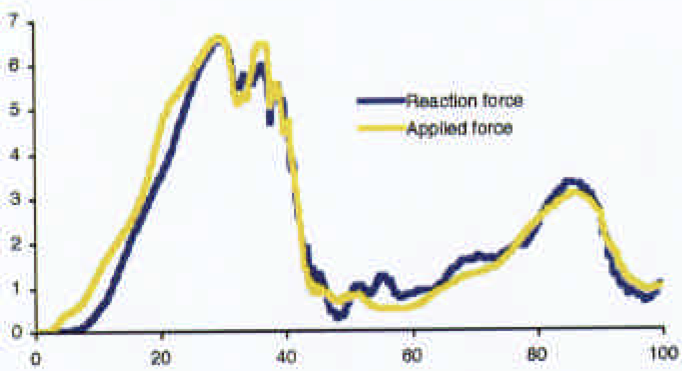


Figure 9. Applied and reaction force histories from a typical test. The similarities between these curves indicate that inertial effects are small because of the relatively low applied loading rate and fixed pelvis boundary condition.

Figure 10 shows neutral-posture force histories from the tests, which indicate that the rise times (or times to fracture) for these tests are within the desired range of 20-60 ms and that the loading rates are all below 300 N/ms. Since the platform impact velocities for all of these tests were similar and inertial effects were small, the differences in observed loading rates are likely due to inter-subject differences in KTH stiffness.

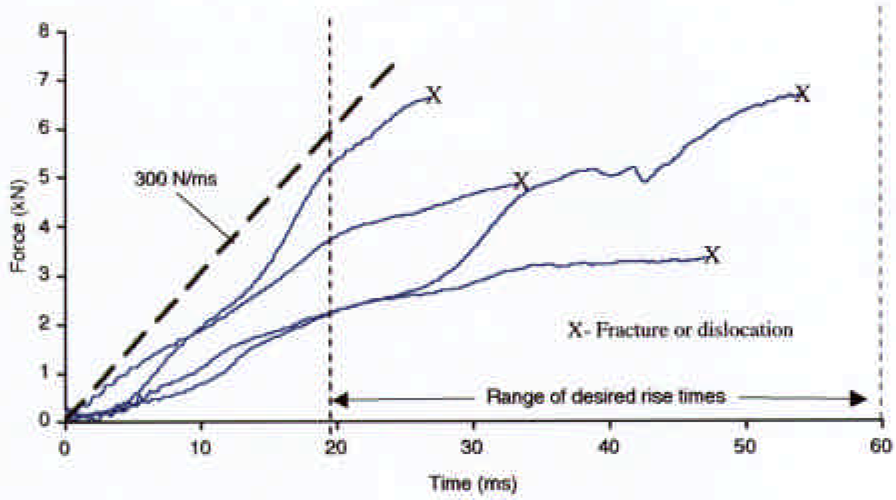


Figure 10. Force histories and times to fracture produced in isolated pelvis testing in the neutral posture. Impact velocities were similar for all tests. Differences in loading rates result from inter-subject differences in KTH stiffness.

When tested under similar loading rates and femur orientations (i.e., flexion and adduction angles), the left and right hips/pelvises of the cadavers used in testing experienced similar patterns of injury at similar forces. The average difference in failure force between left and right hips tested at similar femur orientations is 4%. However, the sample size is not yet large enough to determine if these differences (i.e., the difference between left and right sides and first and second tests) are significant, although the bilateral symmetry of the body and the results from pilot tests imply that it is not significant.

The tolerance of the hip in the neutral posture (i.e., zero flexion, zero adduction) in these tests is $5.2 \pm 1.5$ kN. This is significantly less than the reported tolerances of the knee or femur and suggests that the hip is the weakest component of the KTH complex. For a given posture, the hips of female cadavers in this dataset failed at $4.8 \pm 0.8$ kN, while the hips of their male counterparts failed at $5.6 \pm 0.9$ kN. However, the difference between the male and female hip fracture/dislocation tolerances is not statistically significant ($p = 0.17$).

## CONCLUSIONS

This study is expected to provide the data that are needed to establish the tolerance of the hip joint from loading through the knee and femur. Unlike previous studies that investigated KTH tolerance to knee loading at high loading rates, and thus with high inertial effects, the current study measures hip tolerance in a fixed-pelvis condition, so that the measured hip-tolerance values are almost completely independent of inertial effects.

Injuries produced in the current test series are consistent with real-world hip/pelvis injuries from frontal crashes observed in the U of M CIREN database. Both the fracture patterns and the distribution of hip/pelvis fractures produced in the isolated pelvis tests match the corresponding patterns and injury distribution in the U of M CIREN database.

Additional testing will be conducted to determine the tolerance of the hip over a range of flexion and adduction/abduction conditions. Future work will also investigate how loading rate affects forces produced along the KTH complex. Estimates of loading rate effects will be combined with the tolerance of the hip determined from isolated pelvis testing, as well as femur and knee tolerances, documented in the biomechanical literature, to develop a unified injury prediction model for the entire KTH complex under knee loading.

## ACKNOWLEDGEMENTS

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# DISCUSSION

PAPER: **Development of an Experimental Protocol to Quantify the Tolerance of the Hip to Axial Femur Loading**

PRESENTER: ***Chris Van Ee, UMTRI***

QUESTION: *Guy Nusholtz, Daimler Chrysler*

In one of the graphs you showed the force at the hip and the force at the knee, but there was a very large space lag in the beginning lasting about three or four milliseconds between the two. And if that's to be an inertial load that indicates you're propagating a wave down the femur, and I think the frequency of the time is so long that that wouldn't be possible, the delay propagation. What's the cause of that offset down there?

ANSWSER: I would say that offset is not necessarily a wave, but actually the patella moving back. You first see the inertia of the patella and the tissue on the front and that moves back into the femur so it's more of a stiffness translation there before you pick up and start moving the femur. And as you start to move the femur then you have to get into the pelvis itself and then there's also probably a slight amount of deflection in the gripping actually of the pelvis. So, I think while it's not a stress waive moving down it's more just due to the deflection of each of these portions as they come into contact with each other.

**Q:** So your argument is the laxity? You're saying it takes 1,000 Newtons to move the patella, is that what you're trying to tell me, before you start developing a force back here in the pelvis or are you also including the laxity associated with the acetabulum?

**A:** There's laxity in the hip joint as well as the patella, correct.

**Q:** So that phase leg is due to those two laxities?

**A:** Right. And there's probably laxity in actually gripping the pelvis itself. There are a number of portions where there could be a little laxity.

**Q:** *Jeff Crandall*

Did you get any dislocations at the hip or were they all fractures?

**A:** We did get dislocations.

**Q:** Did I miss it?

**A:** About 15 percent dislocations.