DEVELOPMENT OF INJURY CRITERIA FOR PELVIC FRACTURE IN FRONTAL CRASHES

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ABSTRACT

This paper assesses the position-dependent injury tolerance of the hip in the frontal direction using eight cadaveric subjects. For each subject, the left and right hemipelvis complex were axially loaded using a previously developed test configuration. Six positions were defined from a femur neutral condition, including flexed, neutral, and extended femur positions, and adducted, neutral and adducted positions. Axial injury tolerances based on peak force were found to be 6850 ± 840 N in the extended, neutral position and 4080 ± 830 N in the flexed, neutral position. For the flexed orientation, there is an increase in peak axial force of 18% when the femur is abducted 20^0 and a decrease of 6% when the femur is adducted 20^0. For the extended femur there is a decrease of 4% in abduction axial force and a decrease of 3% in adduction. However, as there is evidence that increases in loading may occur after the initiation of fracture, the magnitude of the peak force is likely related to the extent of injury, not the initial tolerance. Using a potentially more relevant initiation of fracture force value assessed with acoustic crack sensors lowers the injury criteria. The fracture initiation force varied by position from a 3010 ± 550 N in the flexed, neutral position to 5800 N in the extended, neutral position. Further, there was a large position-dependent variation in the ratio of fracture initiation force to the peak axial force. The initiation of fracture was 89% of the peak axial force in the extended, abducted position, but the ratio was 30% in the extended, adducted position. This may have significant implications in the development of pelvic injury criteria and in the design of automobiles to mitigate pelvic injuries.

Keywords: Pelvis, Femur, Hip, Fracture, Injury Criterion, Lower Extremity

THERE ARE APPROXIMATELY 15,000 hip injuries annually in frontal crashes in the U.S (Rupp, et al., 2002). The study of Kuppa and Fessahaie (2003) emphasizes that knee-thigh-hip injuries are the most common of lower extremity injuries and are the most life threatening. Recently, there has been a change noted in the distribution of lower extremity injuries in automobile crashes that increases the prevalence of hip injuries over knee/thigh injuries (Rupp et al., 2003). As hip injuries may be more severe and have greater life years lost than other injuries (c.f. Kuppa et al., 2001, Read et al., 2002) it is important to determine the injury tolerance of the hip to provide injury assessment tools to understand the etiology of injury and develop countermeasures for hip injuries,.

Further, the distribution of hip injuries worldwide varies substantially. In Figure 1, the distribution of lower extremity injuries in frontal crashed with belted occupants in Japan and the United States is reported (Kitagawa, 2003). There is a striking difference in pelvic injuries and thigh injuries in the two countries (Note, proximal femur is included in the thigh results). In Japan, pelvis injuries account for approximately 3% of the lower extremity injuries with thigh injuries accounting for 26% of lower extremity injuries. In contrast, pelvic injuries account for 13% of the lower extremity injuries in the United States with the thigh accounting for 13%. The cause of this change in distribution is unknown, but may be postulated that differences in anthropometry and seating position may play a role. Indeed, Rupp (Rupp, 2004) reports position dependent differences in the femur head/acetabulum contact area. He finds a decrease of 0.6% decrease in load-bearing area in flexion between 0 to 30 degrees flexion from a seated neutral position and 0.9% decrease in load-bearing area in adduction between -15 to 15 degrees of adduction from a seated neutral position. These changes may be further exacerbated by changes in the position dependent profile of bony supporting structures in the hip.

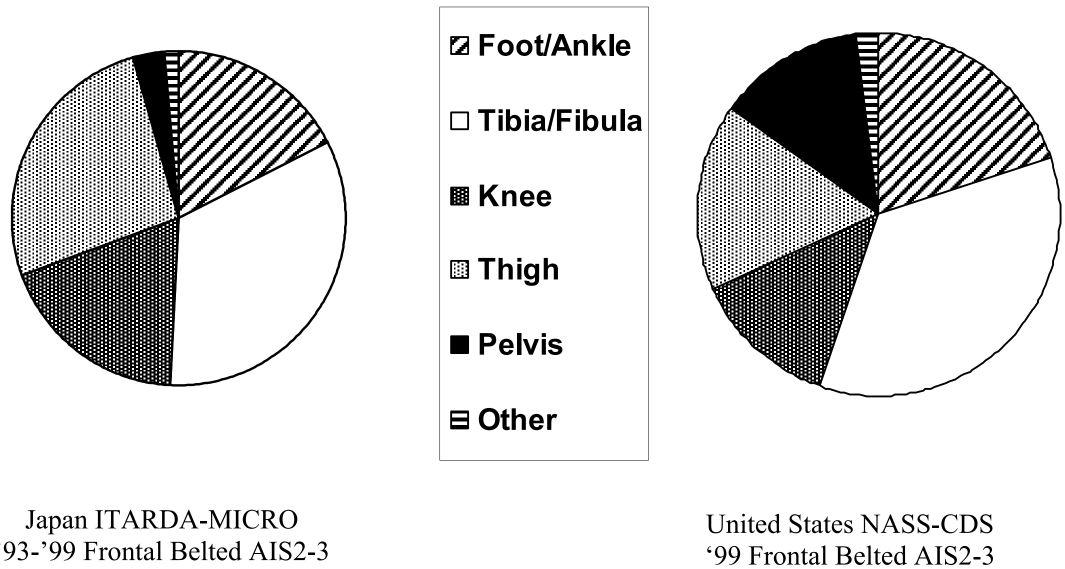


Figure 1. Distribution Of Leg/Pelvis AIS2-3 Injury In Japan And The US (Kitagawa, 2003)

There is a long history of experiments on impacts into the knee of a knee/thigh/hip complex. Such work includes the study of Patrick et al. (1966) in which pelvic injuries occurred to fixed cadaveric specimens from knee loading with 4 to 17 kN of axial force. Nusholtz et al. (1982) used rigid and padded impactors on 16 cadaveric subjects including 37 knee impacts. In these tests, no pelvic or hip fractures occurred at force levels to 37 kN.

Several existing studies utilizing rigid impactors have found no hip injuries to very high force levels (c.f. Powell, et al., 1975, Melvin et al., 1975, Donnelly and Roberts, 1987). For example, Brun-Cassan et al. (1982) performed whole body cadaver tests in which peak knee loading ranged from approximately 4 kN to 11 kN. However, no hip injuries occurred save an iliac crest injury. It has been suggested that the limited number of hip injuries in earlier knee/hip thigh testing arises from characteristics of the experimental loading conditions (Rupp et al., 2003). As many of these tests utilized rigid impactors, it is likely that high impact loading rates and local inertia/failure at or near the knee prevents force transmission to the hip.

More recent work, performed by Yoganandan.et al. (2001), produced hip injuries in an adducted and flexed position from knee loading. However, this work is difficult to use in the determination of injury criterion as the axial forces are unknown. To eliminate the effect of loading rate and inertia on impact tolerance of the knee/hip thigh complex Rupp et al (Rupp et al., 2002, Rupp, et al., 2003) performed a series of tests with knee loading and limited load rates (approximately 300 N/ms). They performed thirty five tests on knee/hip/thigh complexes of 22 cadavers. Of these tests, 25 were performed in a neutral flexion, neutral adduction orientation. Six tests were performed in a neutral and a 10^0 adducted femur position, and four tests were performed in a 30^0 flexed, neutral adduction, abduction position. The flexed and adducted cases were performed as pairs with a neutral test on the contralateral knee/thigh/hip complex. Injury tolerance was assessed in terms of peak axial force at approximately 6.1 kN. Hip tolerance decreased by approximately 34% in the flexed position, and 18% in the adducted posture.

In light of these recent injury tolerance results, and as the relevant injury tolerance regulation in the United States, FMVSS 208, sets a 6805 N limit on mid-femur force for a Hybrid III ATD (FMVSS208, 2003), it is important to further investigate the influence of posture on the injury tolerance of the hip. To eliminate the influence of the knee and distal femur on pelvic tolerance, the current study isolated the pelvis and proximal femur and performed axial loading while limiting the onset rate of loading. The goals of this study are twofold. The first goal is to investigate the position dependent injury criterion for hip injury. The second is to determine if the initiation of bony injury occurs at axial force levels that are below the peak axial forces in the hip.

METHODS

Specimens and Specimen Preparation

Eight cadavers were obtained for pelvic testing by the Center for Applied Biomechanics through the Virginia State Anatomical Board. Cadaver use at the Center for Applied Biomechanics is subject to review by the University of Virginia Cadaver Oversight Committee. Physical parameters of the specimens used in this series are summarized in Table 1. The mean mass of the specimens was 87±15 kg with a mean height 174.5±8.3 cm. The average age at death was 61.5±8.4 years. The specimens chosen were predominantly males to minimize the variation in cadaver force response from osteopenia or osteoporosis. Bone mineral densities shown in Table 1 were obtained using a histogram QCT technique (QBMAPII, The IRIS Inc.). The bone densities of the male specimens was generally good with average bone mineral density of 168 ± 26 g/cm^3 based on the L4 vertebral body. The female specimen, however, had substantially reduced bone density which may explain the low peak axial force to fracture. This specimen was used solely for confirmation of acoustic sensor performance.

The specimens were cryogenically frozen two to three days post-mortem. Radiology reports verified no pre-existing injuries in the specimens. After testing, CT scans and necropsies were performed to determine injury.

Specimen	122	150	153	162	163	171	192	204
Gender	Male	Male	Male	Male	Female	Male	Male	Male
Age at Death	60	67	60	43	65	67	70	60
Body Mass (kg)	78	76	104	113	60	92	NA	75
Stature (cm)	174.2	170.2	177.8	188.0	167.0	185.4	165.1	171.1
Bone Mineral Density (g/cm^3)	NA	152	205	167	122	149	195	140
T – Score	NA	-1.1	0.6	-0.4	-2.5	-1.2	0.3	-1.5
Z - Score	NA	1.3	2.6	0.6	-0.9	1.2	2.9	1.1

Table 1. Specimen Summary (Bone Mineral Densities Based on UCSF Database, T-Scores Based on 25 Year Old Value, Z-Scores Based on Age-Weighted Mean, NA = Not Available)

This series of tests utilized femur/hip/pelvis complexes, comprising primarily the pelvis and sacrum, and approximately 20 to 30 cm of femur. Tissue that did not interfere with the support and connection of the femur to the pelvis was removed. For each test, the crest of the iliac wings and the sacrum was potted in a metal box using syntactic foam, thus providing a rigid mount. The end of each femur was potted in polyester resin in a femur cap, and a femur load cell attached to this cup. The specimen was instrumented with strain sensors (strain gauge rosettes) on the rear of the iliac wing and anterior of the iliac wing. In addition, femur and acoustic sensors were attached to the femur and to the pelvis to determine time of fracture

The potted femur/pelvis complex was attached to a fixture shown in Figure 2 that was based on the fixture reported by Rupp et al. (2002). The test device consisted of a gas-actuated pneumatic impactor that contacts a transfer piston for velocity and energy control. A hydraulic decelerator was used to control the impact force into a load cell mounted to the distal femur of the hemipelvis being tested. The decelerator is programmed to have an onset rate of approximately 400 N/mm, similar to that seen in femur forces during automobile impact testing (UVa Center for Applied Biomechancs, 2003). An attempt was made to provide sufficient energy input into the femur at a constant onset rate to produce fracture in each test. However, two tests produced no fracture despite relatively high force levels.

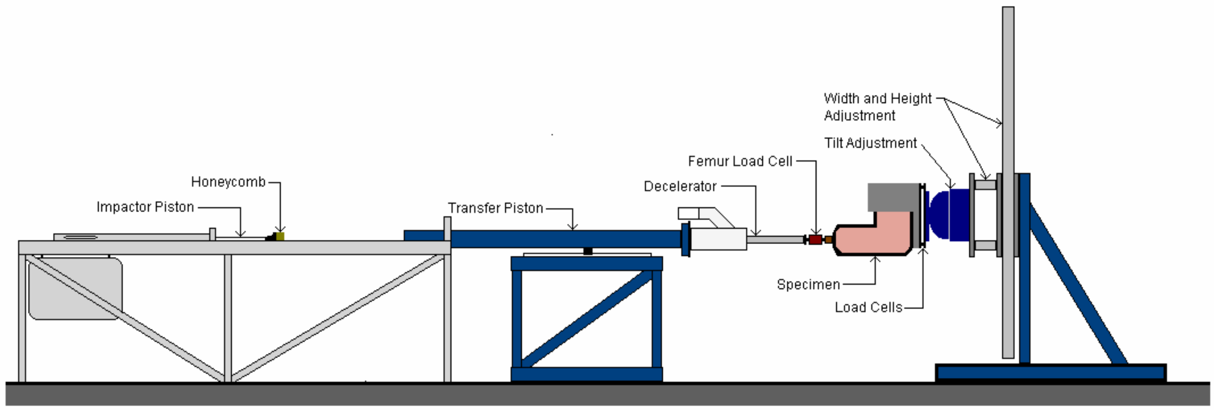
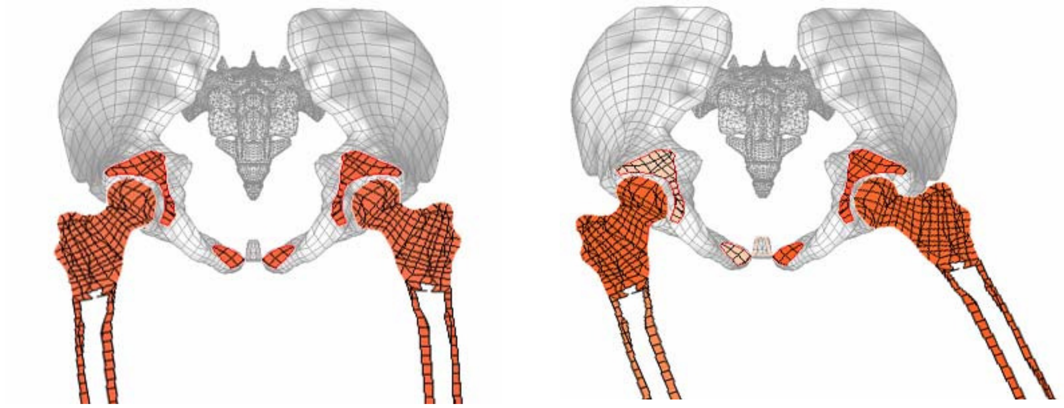


Figure 2. Test Setup.

Positioning and Test Matrix

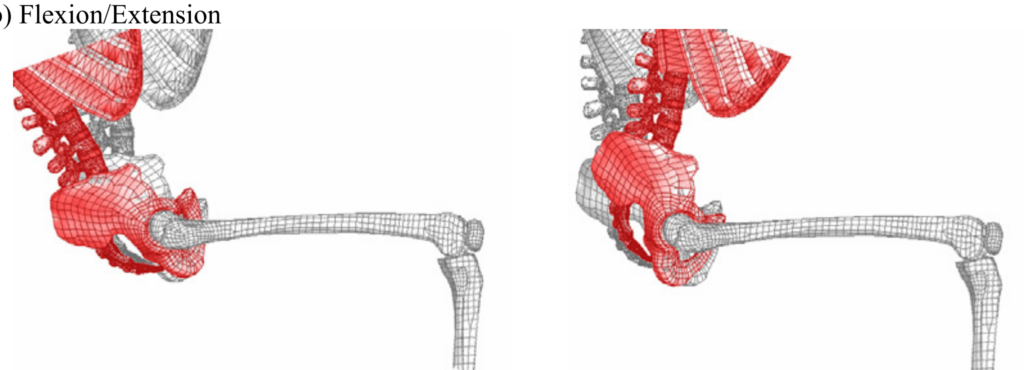
For each specimen, both left and right femur/hemipelvis complexes were tested. The test directions were chosen to represent occupants in frontal crashes. Pelvic ring stability was assessed following each test, and testing did not continue on the contralateral side if stability was compromised. The positions chosen represent a range of driving positions. The postions in adduction/abduction were measured from a femur neutral position, the position in which the mean long axis of the femur is 90^0 from the mid-coronal plane of the torso in both adduction/abduction and flexion/extension. This femur neutral position is aligned with the anatomical neutral in flexion//extension and is 10^0 (on average) from the anatomical neutral in adduction/abduction as shown in Figure 3a. Flexion/extension positions tested range from an extended position 25^0 from femur neutral (representing the position effects of muscle tensing) to a 25^0 flexed position (relaxed). Abduction/adduction positions included adducted (-20^0 from femur neutral), neutral and abducted (20^0 from femur neutral) as shown in Figure 3b.

a) Abduction/Adduction



Anatomical Neutral (Not Tested -10^0 from Femur Neutral) | Femur Neutral | Adducted (-20^0) | Abducted (+20^0)

b) Flexion/Extension



Extended (-25^0) | Flexed (+25^0)

Figure 3. Definition of Pelvis/Femur Complex Positions.

The test matrix for this study is shown in Table 2. Paired tests assess the difference between test conditions for right and left hip/hemipelvis complexes of the same specimen.

	Abducted (-20^0)	**Neutral (0')**	**Adducted (+20^0)**
Flexed (25^0)	122L, 150R, 204R	122R, 162R	162R, 192L
Extended (-25^0)	153R, 150L	153L, 171R	171L, 192R, 204L

Table 2: Test Matrix (Note: The number is the specimen number, and the designation 'L' or 'R' denotes left or right femur/hemipelvis complex.)

RESULTS

Injuries

Eight specimens were tested; there were generally two tests per specimen, left and right, when the pelvic ring was not compromised. Several preliminary tests had artifactural iliac wing fractures that destabilized the pelvic ring. These tests were not included in the analysis. One specimen was used to verify the performance of the acoustic sensors (specimen 163). This specimen, the sole female specimen, saw and AIS 2 fracture at relatively low peak axial load. This is likely the result of lower bone density and smaller bone sizes than the male specimens.

Injuries that occurred during testing can be grouped into four categories: acetabulum fracture (n=10), femoral head/neck fracture (n=3), and dislocations without injury (n=1). Two specimens were not fractured. A summary of injuries is shown in Table 3 with the corresponding AIS scores. Femoral neck fractures occurred only in the extended position. Presumably this results from increased support from the acetabulum causing applied moment to and stress on the femoral neck. In addition, the tests without injury only occurred in the extended position. This is likely due to a similar effect, the increased acetabular support made the injury tolerance of the particular specimen greater than the forces available in the test.

Flex/Ext	Add/Abd	Specimen	Injury	AIS
Flexed	Abducted	122L	comminuted acetabulum	3
		150R	comminuted acetabulum	3
		204R	posterior acetabulum fracture	3
	Neutral	162L	posterior acetabulum fragment	2
		122R	acetabulum, superior ramus fracture	3
	Adducted	162R	Dislocation, disrupted joint capsule	2
		192L	posterior acetabulum fracture	3
Extended	Abducted	153R	posterior acetabulum fracture	3
		150L	none	NA
	Neutral	153L	none	NA
		171R	femoral neck fracture	3
	Adducted	171L	comminuted acetabulum fracture	3
		192R	femoral neck fracture	3
		204L	posterior acetabulum, femoral head fracture	3
Neutral	Neutral[1]	163L	minor acetabulum fracture	2

Table 3. Injury Summary and AIS score (NA=Not Applicable)

Sensor Data

Use of the hydraulic decelerator allowed a relatively consistent axial loading rate range with a mean of 370±120 N/ms as shown in Figure 4. The highest loading rate seen in this series was 540 N/ms and lowest was 170 N/ms. These loading rates are similar to those seen in dummy femur forces in automobile crash testing (Rupp et al., 2003, UVa Center for Applied Biomechanics, 2003).

Since the pelvis was fixed, the impactor continued positive displacement after the time of fracture and peak force although the maximum displacement was limited. The primary measurand was femur axial force; both peak axial femur force and axial femur force at fracture are shown in Table 4. The peak forces that occurred during the tests varied with position as shown in Figure 5. Generally, the variation followed the variation in support area of the acetabulum on the femoral head. As discussed above, the support area of the acetabulum increases with extension and decreases with flexion. This support area also increases with adduction and decreases with adduction. So, the tests with the most structural support, the abducted, extended cases, had large peak forces (6600 ± 1300 N). In contrast, the test position with the least structural support from the acetabulum, the adducted, flexed, case saw lowest peak forces (3800 ± 200 N).

Though there were a limited number of samples in each test condition, the peak forces were found to be 6850 ± 840 N in the extended, neutral position and 4080 ±830 N in the flexed, neutral position. For the flexed orientation, there is an increase in peak axial force of 18% when the femur is abducted

[1] Specimen used for verification of acoustic sensor performance. Femur/hemipelvis specimen 163R was not tested.

20^0 and a decrease of 6% when the femur is adducted 20^0. For the extended femur there is a decrease of 4% in abduction axial force and a decrease of 3% in adduction.

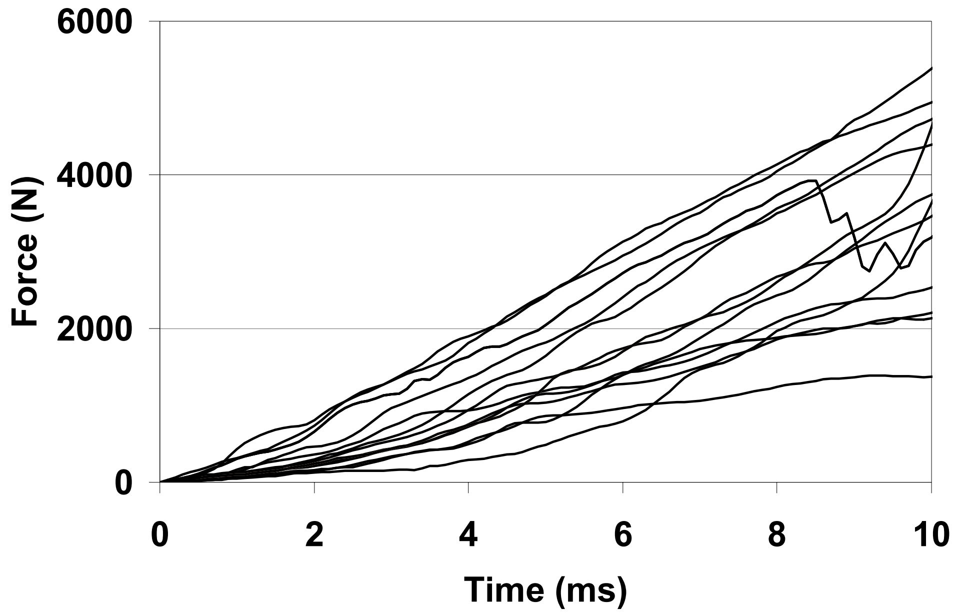


Figure 4. Axial Femur Force Time Histories.

Specimen	Flex/Ext	Add/Abd	Fracture	Peak Axial Femur Force (N)	Axial Femur Force at Fracture (N)
122L	Flexed	Abducted	Y	4377	2003
122R	Flexed	Neutral	Y	3490	3404
150L	Extended	Abducted	Y	7481	NA
150R	Flexed	Abducted	Y	4519	DF
153L	Extended	Neutral	N	7450	NA
153R	Extended	Abducted	Y	5664	5466
162L	Flexed	Neutral	Y	4664	2615
162R	Flexed	Adducted	Y	4348	3356
163L	Neutral	Neutral	N	150	NA
163L	Neutral	Neutral	N	1192	NA
163L	Neutral	Neutral	Y	1808	1671
171L	Extended	Adducted	Y	7413	1536
171R	Extended	Neutral	Y	6255	5821
192L	Flexed	Adducted	Y	3316	3146
192R	Extended	Adducted	Y	4634	1793
204L	Extended	Adducted	Y	7953	2734
204R	Flexed	Abducted	Y	5526	4204

Table 4: Peak and Fracture Axial Force (Note: DF = Data Acquisition Failure, NA=Not Applicable)

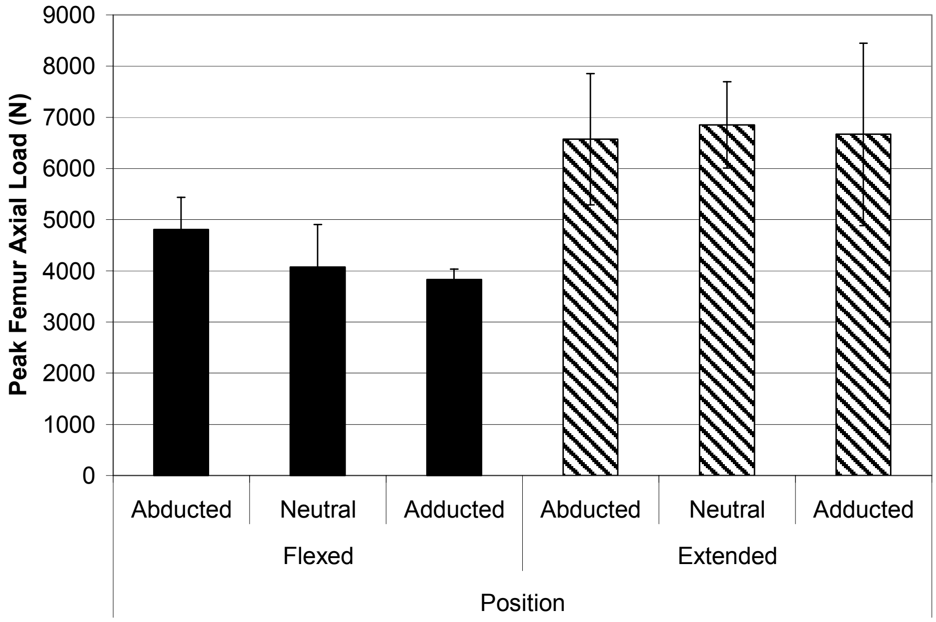


Figure 5. Peak axial femur force vs. position (Error bars are ±1 standard deviation).

However, there is evidence from the testing that the peak forces may be substantially higher than the forces necessary to initiate fracture. For example, an elongated force plateau is seen in the force trace for the test (specimen 162R) with a force of approximately 3500 N (Figure 6). At the onset of this ramp, an acoustic burst consistent with fracture is seen. This result is similar to that seen by Funk in axial lower extremity testing (Funk, 2001) and likely represents cracking in the femoral head and joint capsule. After the force plateau, the force continues to build to over 4000 N.

As the structure of the femoral head/acetabulum interface is complex, there is no reason to suspect that the axial force to initiate fracture is a fixed percentage of the peak axial force. Indeed, from the limited number of samples in the current study, the radio of fracture force to peak axial force varied substantially from extended, adducted (70% difference) to flexed, abducted (15% difference) (Figure 7). These differences may be attributed to the initial failure of bony structures under position dependent stress concentrations. Once these structures (such as the acetabular lip) fail, the stress may be further distributed, and axial forces may increase.

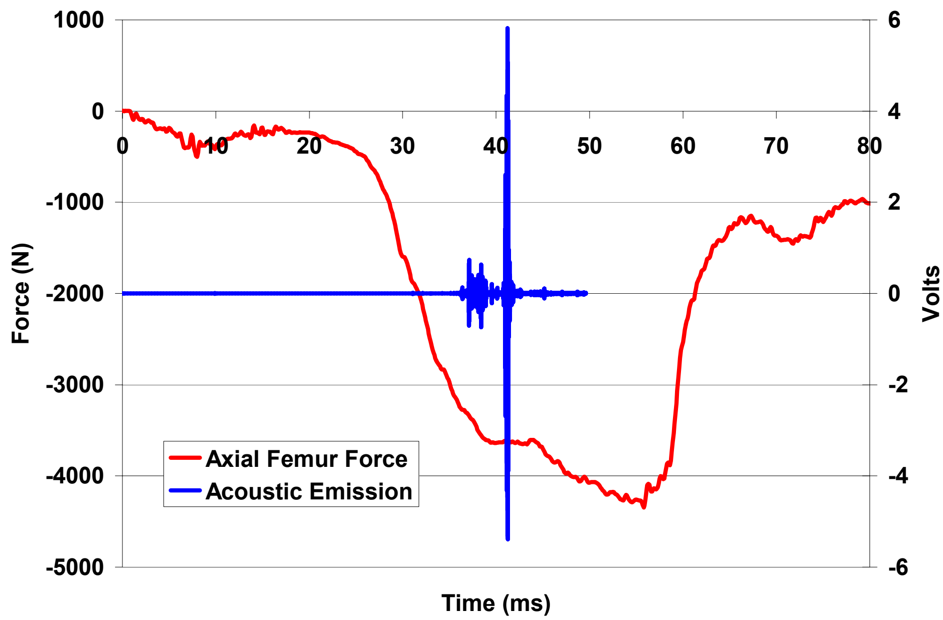


Figure 6. Sample Axial Force Time History With Corresponding Acoustic Signal (Specimen 162R)

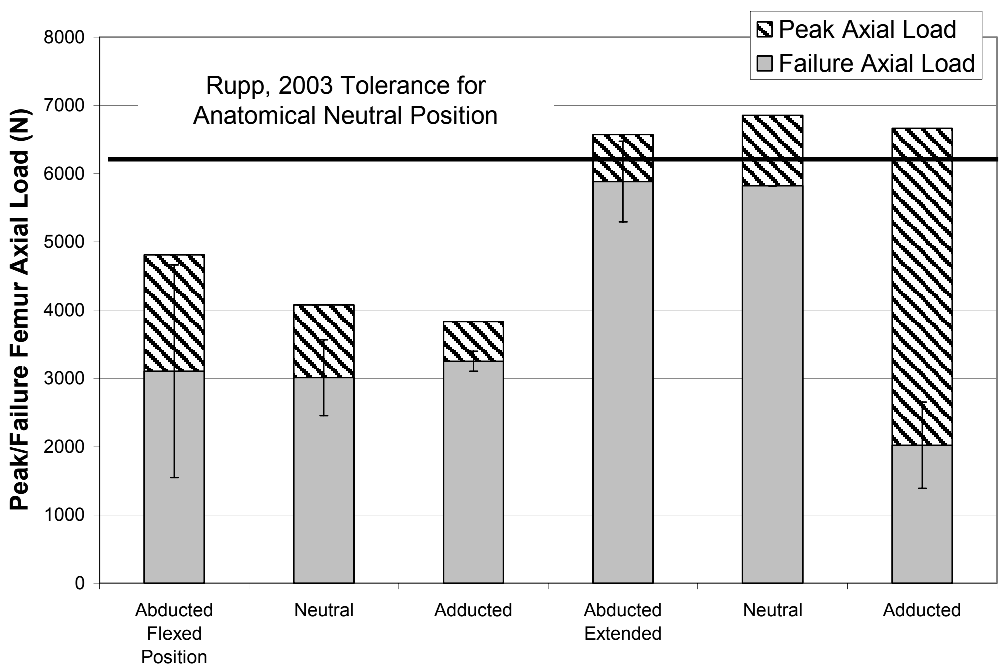


Figure 7. Peak/failure axial femur force vs. position. Solid bars are failure force, crosshatched bars are peak force (Error bars are ±1 standard deviation).

To verify the results of the acoustic sensors, one test specimen (163) was used with multiple levels of impact force to correlate presence of nonzero acoustic sensor data with macroscopic (palpable) fracture. As shown in Figure 8, the acoustic sensor traces from tests with 150 and 1200 N peak force showed no indication of fracture in the acoustic sensor traces while the test with 1800 N peak force had bursts typical of bone fracture. In addition, the final test had a palpable fracture of the acetabulum suggesting a correlation of bony fracture with the acoustic signal.

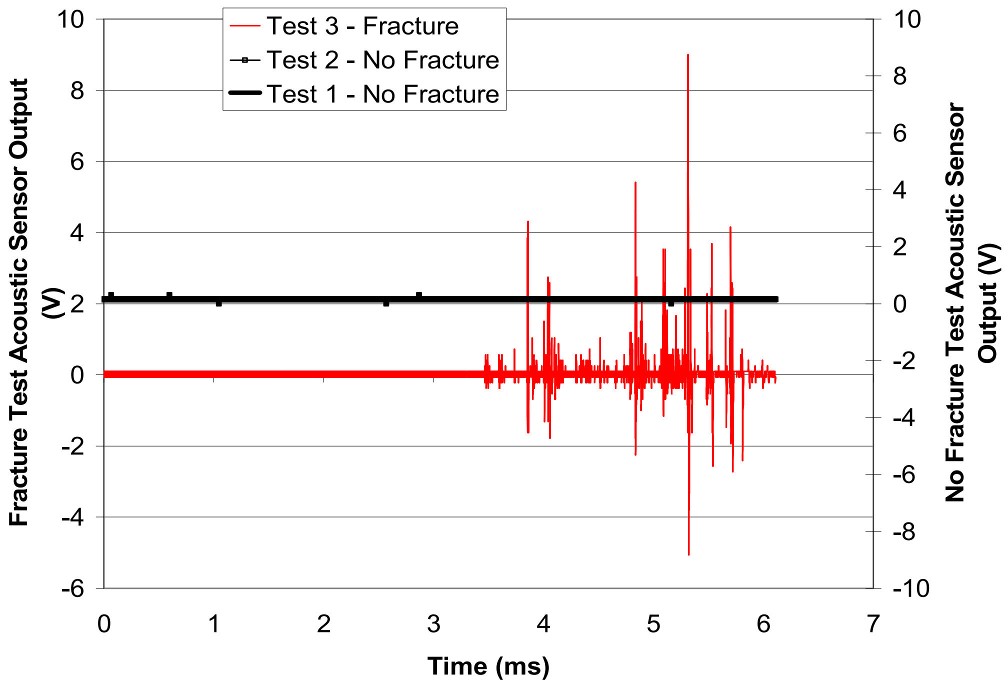


Figure 8. Acoustic Sensor Output for Fracture/No Fracture Tests with Repeated Loading (Specimen 163L).

DISCUSSION

Though detailed positions of the current study and those of Rupp et al. (2003) were different, peak axial forces obtained in this work are similar to those seen by Rupp *et al* as shown in Figure 7. Rupp et al. found an average peak axial force of 6.1 kN in the neutral posture, while the current study sees approximately 6800 N in the femur neutral, neutral position. Rupp found generally decreased peak forces in adduction and flexion, similar to this work. Rupp saw decrease of 34% in flexion of 30^0 from an anatomical neutral position, while the current study saw a decrease of 40% from 20^0 extended to 20^0 flexed Further, Rupp et al., obtained an 18^0 decrease in peak axial force from the anatomical neutral position to 10^0 adducted from the neutral posture. In the current study, there was a decrease of 6% in peak axial force from femur neutral to abducted 20^0 from femur neutral.

The use of peak forces in the femur or knee as the basis for an injury tolerance may be questioned. There is evidence that increases in loading may occur after the initiation of fracture, so that the magnitude of the peak force is likely related to the extent of injury, not the initial tolerance. Using a potentially more relevant factor, the initiation of fracture force value assessed with acoustic crack sensors, lowers the injury tolerance, in some cases substantially. It seems prudent that further samples be tested to elucidate the relationship between the axial fracture force, the axial peak force, and the position of the femur relative to the pelvis.

CONCLUSION

The position dependent injury tolerance of the hip in frontal crashes was investigated in this study. Though a limited number of specimens was available for testing, peak axial forces were found to be 6850 ± 840 N in the extended, neutral position and 4080 ±830 N in the flexed, neutral position. For the flexed orientation, there is an increase in peak axial force of 18% when the femur is abducted 20^0 and a decrease of 6% when the femur is adducted 20^0. For the extended femur there is a decrease of 4% in abduction peak axial force and a decrease of 3% in adduction. The difference in peak forces may be attributed to differences in the structural support on the femoral head from the acetabulum. For example, the test with the most acetabular structural support (extended, abducted) saw the second highest peak forces (6600±1300 N) while the test condition with the least structural support (flexed, adducted) saw the lowest peak forces (3800±200 N).

However, as there is evidence that increases in loading may occur after the initiation of fracture, the magnitude of the peak force is likely related to the extent of injury, not the initial tolerance. Using a potentially more relevant initiation of fracture force value assessed with acoustic crack sensors lowers the injury criteria. The fracture initiation force varied by position from 3010±550 N in the flexed, neutral position to 5800 N in the extended, neutral position. Variation in fracture initiation force from the neutral adduction/abduction position could be substantial. The percentage decrease from the peak axial force varied from 11% in the extended, abducted position to nearly 70% decrease in the extended, adducted position. Owing to the complexity of the loading and load bearing surfaces, both pre-fracture and post-fracture, there is no expectation that the ratio of force at onset of fracture and peak force is constant. Indeed, the peak axial force substantially overpredicts the axial force to initiate fracture in most cases. This may have significant implications in the development of pelvic injury criteria and in the design of automobiles to mitigate pelvic injuries.

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REFERENCES

Brun-Cassan, R., Leung, Y., Tarriere, C., Fayon, A., Patel, A., Hureau, J., Determination of knee-femur pelvis tolerance from the simulation of car frontal impacts. Proceedings of the Seventh International Conference on the Biomechanics of Impacts, 1982.

Donnelly, B.R., and Roberts, D.P., Comparison of cadaver and Hybrid III dummy response to axial impacts of the femur, Proceedings of the Thirty-First Stapp Car Crash Conference, Society of Automobile Engineers, Warrendale, PA, 1987.
FMVSS-208 (Federal Motor Vehicle Safety Standards), Code of Federal Regulations, 49CFR571.208S15.3.5, 2003.

Funk, J., PhD Dissertation, University of Virginia, 2001.

Kitagawa, I., Personal Communication, 2003.

Kuppa, S., and Fessahaie, O., An overview of knee-thigh-hip injuries in frontal crashes in the United States, Proceedings of the 18th Conference on the Enhanced Safety of Vehicles, Paper 416, United States Department of Transportation, Washington, D.C, 2003.

Kuppa, S., Wang, J., Haffner, M., and Eppinger, R., Lower extremity injury and associated injury criteria, Proceedings of the 17th Conference on the Enhanced Safety of Vehicles, Paper 457, United States Department of Transportation, Washington, DC, 2001.

Melvin, J.W., Stalnaker, R.L., Alem, N.M., Benson, J.B., and Mohan, D., Impact response and tolerance of the lower extremities, Proceedings of the Nineteenth Stapp Car Crash Conference, Society of Automobile Engineers, Warrendale, PA, 1975.

Nusholtz, G., Alem N., and Melvin, J., Impact response and injury to the pelvis, Proceedings of the Twenth-sixth Stapp Car Crash Conference, Society of Automobile Engineers, Warrendale, PA, 1982.

Patrick, L., Kroell, C. and Mertz, H., Forces on the human body in simulated crashes, Proceedings of the Ninth Stapp Car Crash Conference, 1966.

Powell, W.R., Ojala, S.J., Advanti, S.H. and Martin, R.B., Cadaver femur responses to longitudinal impacts, Proceedings of the Nineteenth Stapp Car Crash Conference, Society of Automobile Engineers, Warrendale, PA, 1975.

Rupp, J.D., Reed, M.P., Van Ee, C.A., Kuppa, S., Wang, S.C., Goulet, J.A., and Schneider, L.W., The tolerance of the human hip to dynamic knee loading, Stapp Car Crash Journal, vol. 46, pp. 211-228, 2002.

Rupp, J.D., Reed, M.P., Jeffreys, T.A.., and Schneider, L.W., Effects of Hip Posture on the Frontal Impact Tolerance of the Human Hip Joint, Stapp Car Crash Journal, vol. 47, pp.21-33, 2003.

Rupp, J.D., Personal Communication, 2004.

University of Virginia Center for Applied Biomechanics, Unpublished data, 2003.

Yoganandan, N.A., Pkintar, F.A., Genneralli, T.A., Maltese, M.R., and Eppinger, R.H., Mechanisms and factors involved in hip injuries during frontal crashes, Stapp Car Crash Journal, vol. 45, pp. 437-448, 2001.