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Bone Density and Trochanteric Tissue Thickness Affect Fracture of the Female Pelvis in Side Impact Experiments

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ABSTRACT

In spite of advances in occupant protection, pelvic fractures remain a source of morbidity and mortality in motor vehicle side impacts. Women are more frequently injured than men, leading automotive researchers to incorporate the SID-IIs (small female crash dummy) into side impact testing of passenger vehicles. Currently accepted fracture tolerances for the female pelvis are based on extrapolations from experimental results containing both men and women, and there is no unified theory that predicts pelvic fracture for a woman of a given body habitus under side impact conditions. The present report describes a two phase experimental study aimed at improving our understanding of pelvic fractures, with an emphasis on the roles of bone quality and soft tissues. The first phase, which explored the role of bone mineral density (BMD) on impact responses, involved drop tower tests on isolated bone-ligament specimens, both male and female. The second phase employed a linear impactor and intact specimens (females only) positioned on a sliding, inertially compensated seat, where the influence of trochanteric soft tissue thickness (T) along with BMD was studied. Positive correlations between the force to fracture the pelvic ring and BMD were observed in both sets of experiments. For the intact female specimens of Phase 2, the force to fracture correlated positively with BMD and T; while impulse correlated with T only. Maximum compression, viscous criterion, and energy to fracture, however, were independent of both BMD and T. As anticipated, the force to fracture an intact pelvis was greater than that of an isolated bone-ligament specimen, for a given BMD. The force tolerance at 25% probability of fracture for the intact female specimens (average age = 76) was 3.16 kN, which is substantially below previously reported estimates for the 5th percentile female.

INTRODUCTION

Motor vehicle crashes are the most common cause of serious injury (US DHHS, 2000). Moffat et al. (1990) estimated that, in 1985 alone, 15,300 pelvic fractures occurred during motor vehicle crashes in the United States. Since the early 1990's, a significant increase in the incidence of pelvic fractures has been documented, perhaps as a result of improved overall survival rates (Inaba et al., 2004). Side impacts are the most common cause of pelvic fractures, for which the mortality rate may be as high as 50% due to concurrent intra-abdominal hemorrhaging (Gokcen et al., 1994). These collisions tend to produce lateral compression (LC) fractures of the pelvic ring that involve the pubic rami and acetabulum (LC-I), and with increased severity, central dislocation of the femoral head (Grattan and Hobbs, 1969; States and States, 1968) and/or separation at the pubic symphysis and sacroiliac joints (LC-II, LC-III) (Tile and Hearn, 1995; Siegel et al., 1993).

The Abbreviated Injury Scale (AIS) is commonly used to classify the severity of injury, where closed (stable) pelvic fractures (with or without hip dislocation) are scored as AIS = 2, while open, displaced, or comminuted fractures (unstable) are scored AIS = 3 (Abbreviated Injury Scale, 1998). One clinical study found that 100% of victims of unstable pelvic fractures and over 50% of patients recovering from stable pelvic fractures experienced chronic pain (Mkandawire et al., 2002). Thomas and Frampton (1999) determined that fractures of the pelvis scored higher than any other form of injury on the HARM scale (Malliaris et al., 1985), which reflects economic costs associated with long-term outcome.

Women appear to suffer the majority of pelvic fractures (Seigel et al., 1993), with injury rates up to three times that of men (Rowe et al., 2004). Such findings led the Insurance Institute for Highway Safety (IIHS) Vehicle Research Center to begin using the SID-IIs crash dummy (5^{th} percentile female: 5'1", 106 lb.) in its crash performance testing. Results suggest that the SID-IIs will be more effective in the promotion of side impact protection (Arbelaez et al., 2002). This necessitates accurate experimental correlations of fracture forces and accelerations in women and how they relate to pelvic loads measured in dummy tests.

Current understanding of pelvic fracture tolerances in side impacts is based primarily on cadaver testing done between the late 70's and early 90's. The earliest studies defined tolerances in terms of critical values of acceleration (62-120g) (Haffner, 1985; Marcus et al., 1983; Kallieris et al., 1981; Tarriere et al., 1979; Fayon et al., 1977) leading to Federal Motor Vehicle Safety Standard 214. This standard mandates the maximum allowable sacral acceleration of 130g, for a 50^{th} percentile male, in a 33.5 mph side impact crash. Recently, an amendment to FMVSS 214 has been proposed that utilizes the SID II-s dummy in an oblique pole impact, as well as the current moving deformable barrier side impact tests (NHTSA, 2004). The proposed injury tolerance criterion (pelvic force: iliac wing + acetabulum load cells = 5.3 kN), however, was developed by scaling experimental data that included primarily males (Bouquet et al., 1998).

One confounding difficulty in cadaver testing is the anthropometric variability between cadaver specimens. A variety of scaling techniques have been utilized in order to compensate for variation in body habitus, including the Livi Index (Cesari and Ramet, 1982) and the Lambda ratio (Eppinger et al., 1984). Cesari et al. (1982, 1980) impacted full cadavers and found a fracture force for the 50^{th} percentile male of 10 kN. From their data, which included both males and females, they extrapolated a 4 kN critical impact value for the 5^{th} percentile female using the Livi index. Cavanaugh et al. (1993), on the other hand, reported a 5.77 kN tolerance for the 5th percentile female using the Lamda ratio to scale data from 17 tests with 7 females. The basis for the pelvic tolerances for the SID-IIs in the proposed amendment to FMVSS 214 (Bouquet et al., 1998) was developed by scaling a 7.6 kN general tolerance from 31 tests including only 5 females using the Lambda ratio. Viano (1989) showed that lateral compression correlated well with pelvic fracture, while impact force and acceleration did not. Lau & Viano (1986) proposed new injury tolerance measures called the "viscous criteria," which involved products of impactor velocity and pelvic compression, $(VC)_{max}$ and $V_{max}C_{max}$. Cavanaugh et al. (1990) observed that compression and $V_{max}C_{max}$ were the best predictors of pelvic fracture in their tests.

Neither the Livi index nor Lambda ratio directly accounts for variations in bone quality. In the last decade, studies have demonstrated that reduced bone mineral density (BMD) was associated with lower fracture loads in isolated cadaveric femurs (Pinilla et al., 1996; Bouxsein et al., 1995; Courtney et al., 1994). This led us to conjecture that the human pelvis would behave similarly under side impact conditions. The first phase of the present report (Beason et al., 2003) was performed to determine the role of hip BMD on pelvic fracture. Drop tower experiments were performed on isolated bone-ligament specimens and the relationship between fracture forces and hip BMD was determined. It was hypothesized that the force to fracture the pelves would correlate with hip BMD. Such information may ultimately be useful for improving protection of the pelvis for persons of varying ages and genders, which are associated with differences in BMD (Ashton-Miller and Schultz, 1997).

Recent studies revealed lower AIS scores for obese occupants verses non-obese in side impacts (Arbabi et al., 2003) and that increased levels of subcutaneous fat create a "cushion" that may decrease injury severity in women (Wang et al., 2003). Furthermore, trochanteric soft tissues have been shown to attenuate forces and increase energy absorption during experimental falls (Robinovitch et al., 1995). The second phase of the present study (Etheridge et al., 2005) employed a linear impactor, added trochanteric soft tissue thickness as an independent variable, and included only female specimens. It was hypothesized that the force to fracture the pelves would correlate positively with total hip BMD and trochanteric tissue thickness. Furthermore, it was conjectured that the force to fracture the intact pelves in Phase 2 would be higher than the force necessary to fracture the isolated pelves in Phase 1. The present experiments were intended to establish improved injury tolerances for pelvic fracture in women in side impacts, which might ultimately be used to better interpret SID-IIs forces measured during vehicle side impact crash tests.

METHODS

Phase 1 – Drop tower tests on bone-ligament specimens

Twelve fresh-frozen cadaveric pelves (age = 68 ± 16 years), sectioned at the fourth lumbar vertebra and proximal femurs, and free of pre-existing fractures, were obtained through the University of Alabama at Birmingham (UAB) Willed Body Program, as approved by the UAB Institutional Review Board (IRB). When available, age, gender, height, and weight were documented for each cadaver donor (Table 1). The pelves were dissected of muscle and adipose tissues, while the ligaments spanning the sacroiliac joints and pubic symphysis were left intact. To assess bone quality, total hip BMD was measured by dual energy x-ray absorptiometry (DEXA, QDR 4500, Hologic Inc., Waltham, MA) for each specimen (Figure 1). The isolated pelves were stored at -20°C until 24 hours prior to impact preparations, at which time they were thawed to 4°C. Specimen preparations for each impact were conducted over a period of several hours, which allowed the pelves to further acclimate to room temperature. Saline solution was applied periodically to the pelvic bones and ligaments to keep them moist during preparation and testing.

The remnant vertebral column was potted in polymethyl methacrylate (PMMA). The pelves were secured in a fixture designed to hold the isolated pelvis on its side while providing normal and frictional support at the ischial tuberosities (Figure 2). To simulate being in a seated position, a compressive pre-load (65% body weight) was applied along the longitudinal axis of the vertebral column (Molz et al., 1997). Breakaway wire was used to hold the impact-side femur in 90 degrees flexion with neutral abduction/adduction and internal/external rotation. The contralateral (left) iliac wing and greater trochanter were fixed in PMMA similar to previous work, which had shown that artifactual fractures were not caused by the rigid support conditions (Guillemot et al., 1997).

Table 1. List Of Cadaveric Specimens And Known Characteristics For Phase 1.

Pelvis	Age	Sex	Weight (lb)	Height (in)	BMD (gm/cm^2)
#43[(1)]	68	M	157	74	0.82
#56	70	M	185	73	1.24
#57	72	M	175	71	0.61
#58	61	M	180	72	0.90
#59	62	M	180	70	0.92
#63	60	F	--	--	0.64
Mean ± S.D.	65.5 ± 5.13		175 ± 10.9	72.0 ± 1.57	0.86 ± 0.23
#60[(2)]	62	M	165	69	0.73
#61	27	M	145	67	0.97
#75	78	F	--	66	0.76
#76	86	F	132	--	0.68
#77	81	F	120	65	0.17
#78	85	F	109	62	0.73
Mean ± S.D.	69.8 ± 22.7		134 ± 21.8	65.8 ± 2.60	0.67 ± 0.27

[1]Unpadded group.
[2]Padded group.

Figure 1: Example DEXA scan showing trace for determination of total hip BMD that includes the acetabulum, femoral head and neck, and the greater and lesser trochanters.

Six reflective pin markers (5-mm diameter) were fixed to specific locations around the inner portion of the pelvic ring (Figure 2). The two-dimensional coordinates of the markers were captured with a ProReflex MCU 1000 (1 kHz) high-speed infrared camera (Qualisys Inc., Glastonbury, CT). The camera was positioned normal to the plane of reflective markers. Relative marker displacements were used to monitor fractures of the pelvic ring during the impact event.

Lateral impacts to the greater trochanter were performed with a DYNATUP™ 8250 drop tower (Instron Corporation, Canton, MA), which was equipped with a drop mass, load cell, and aluminum impactor (3″ × 4″ striking area) (Figure 2). A drop height of approximately 1 m produced a mean impact velocity of 4.49 ± 0.03 m/s, which previously produced fractures in some pelves and no fracture in others (Molz et al., 1997). Force data were recorded with the GRC 830-I data acquisition system (GRC International, Inc., Santa Barbara, CA) at a sampling rate of 20.5 kHz. A velocity trigger measured the impact velocity, V_i, of the total falling mass. Inertial compensation was performed according to Bouquet et al. (1998, 1994), where the ratio of the total mass (m = 13.4 kg) to the mass above the load cell (12.9 kg) was computed to be 1.04. This factor was used to scale the force measured by the load cell, $p(t)$, through the equation,

$$F(t) = mg - 1.04p(t), \tag{1}$$

where $F(t)$ is the inertially compensated total force and g is acceleration due to gravity (9.81 m/s^2). $F(t)$ was then used to calculate the time-dependent acceleration of the impactor, $a(t)$, which was then integrated to obtain velocity, $V(t)$, and deflection, $y(t)$. The energy dissipated during the impact, $E(t)$, was calculated using the equation,

$$E(t) = m/2\ (V_i^2 - V^2(t)) + mg\ y(t). \tag{2}$$

Energy to peak load, E_{peak}, was defined as the dissipated energy up to the time to peak load, t_{peak}.

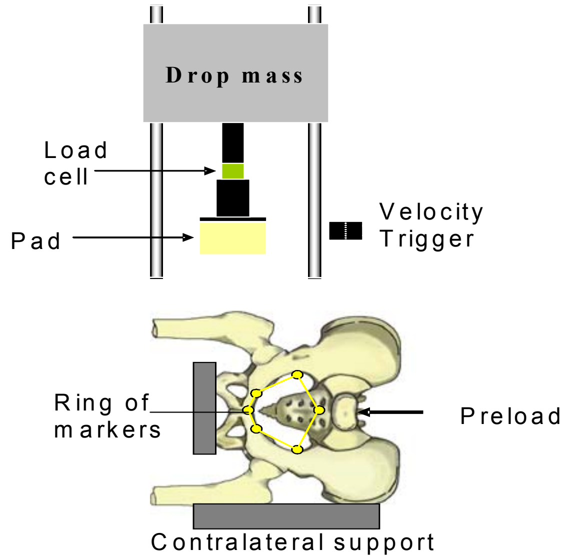

Figure 2: Schematic of drop tower impacts. Isolated bone-ligament specimens were positioned on their side with femurs flexed at approximately 90 degrees, potted contralaterally in PMMA, compressively preloaded through the spine and supported with frictional restraint at the ischial tuberosities.

Six pelves were impacted without any padding between the impactor and bone. Six other pelves were impacted with a flat 56-mm thick piece of 32 kg/m^3 polyethylene foam (Atlas Foam Inc., Sylmar, CA) added to the impactor. This padding material was selected for its superior energy absorption capabilities found in pilot thesis work (Arbelaez, 1999). Following the impacts, the pelves were scanned with a GE LightSpeed QX/i CT scanner (General Electric Medical Systems, Milwaukee, WI). The CT scans were examined to identify any fractures resulting from the impacts.

Total hip BMD was tested for correlation with impact response parameters (F_{max}, E_{peak}, t_{peak}) for fractured pelves using linear regression. For all statistical tests, α = 0.05 was designated as the level of statistical significance. F_{max} and t_{peak} were also used to compare our impact conditions to earlier full cadaver studies (Zhu et al., 1993; Cavanaugh et al., 1990; Viano, 1989; Cesari and Ramet, 1982; Nusholtz et al., 1982; Cesari et al., 1980).

Phase 2 – Pneumatic lateral impacts of intact, seated specimens

Ten fresh-frozen specimens from female cadavers (age = 76 ± 9 years) were received sectioned at the fourth lumbar vertebra (L4) and proximal femurs, as approved by UAB IRB. Age, height, and weight were recorded (Table 2). The pelves were stored at -20°C until approximately 48 hours prior to testing at which time the specimens were allowed to thaw and acclimate to room temperature. In addition to the DEXA scan for total hip BMD as done in Phase 1, computed tomography scans (CT, LightSpeed QX/i, GE Medical Systems, Milwaukee, WI) were taken of each torso in a supine position. Trochanteric soft tissue thickness, T, was averaged from five slices (1 mm thickness, transverse plane) on which the greater trochanter was visible (Figure 3, Table 2).

Table 2. List of cadaveric specimens and known characteristics for Phase 2.

Pelvis	Age	Sex	Weight (lbs)	Height (m)	BMD (gm/cm^2)	T (cm)
#83	74	F	150	1.65	0.60	5.18
#84	76	F	140	1.68	0.78	2.94
#85	82	F	110	1.65	0.84	3.78
#86	82	F	105	1.63	0.45	7.90
#88	78	F	125	1.65	1.02	3.68
#89	82	F	130	1.65	0.66	4.46
#90	78	F	130	1.63	0.74	1.34
#91	53	F	200	1.60	0.84	5.96
#92	74	F	125	1.68	0.53	3.72
#93	80	F	110	1.68	0.87	2.32
Mean ± S. D.	75.9 ± 8.62		132.5 ± 27.5	1.65 ± 0.03	0.73 ± 0.17	4.13 ± 1.88

As done in phase 1, the L4 vertebra was potted in PMMA, which offered a point of contact for the vertebral pre-loader, which was now attached to the back of a custom-built seat and used to apply compressive force equal to the upper body weight (Figure 4). Each pelvis was belted into the seat, which incorporated a contralateral support and permitted the addition of masses to increase the overall weight to that of the original full cadaver (± 5 lbs). The seat was mounted on linear guide rails via ultra low friction bearings that allowed lateral translation of the seat, preloader, and pelvic specimen as one unit during the tests. Sylgard 527 gel (Dow Corning, Midland, MI, USA) was placed between the iliac wings to simulate the missing abdominal viscera (Molz et al., 1997).

Figure 3: Example CT scan demonstrating measurement of trochanteric soft tissues for the pelvis lying supine.

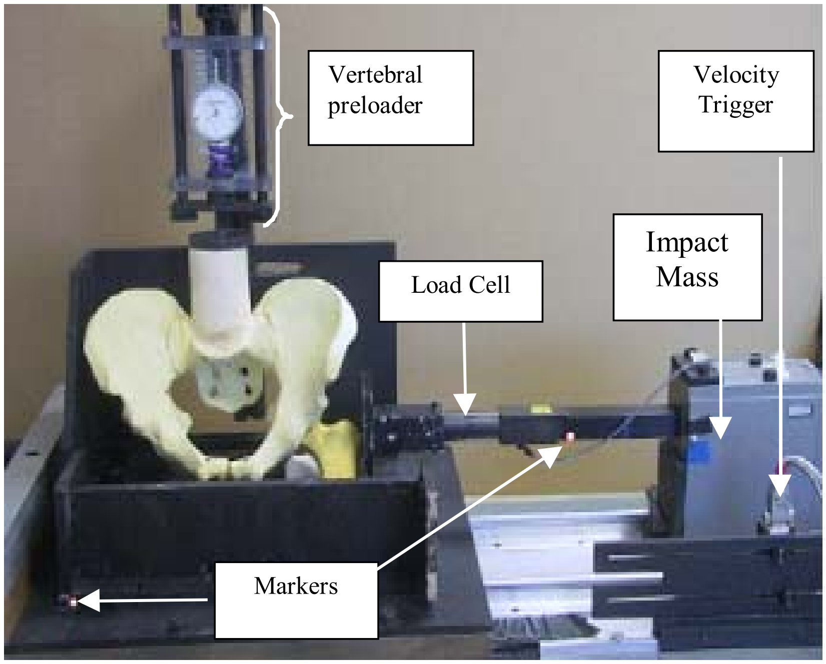

Figure 4: Pneumatic impact test set-up. Each pelvis (plastic model shown) was seated and constrained as shown, including compressive pre-load and a contralateral rigid plate. The impact mass struck the greater trochanter. Impact force was measured with the load cell, while pelvic compression was obtained from high-speed films of marker motion.

The pelves were subjected to lateral impacts centered on the left greater trochanter using a horizontal linear pneumatic impactor (VIA Systems, Brighton, MI, USA). A compressed air reservoir and pneumatic cylinder accelerated a 22.1-kg impact mass with a 3” x 4” flat aluminum striker to impact velocities similar to previous experiments (Bouquet et al., 1998; Viano, 1989; Cesari and Ramet, 1982). The first six specimens (#83-86, 88, 89) were impacted initially at a velocity of 3.31 ± 0.07 m/s. Upon radiographic examination, no fractures were observed. The pelves were struck again at a higher velocity (6.42 ± 0.21 m/s) and subsequent CT scans confirmed that each had fractured. The four remaining pelves (#90, 91, 92, 93) were impacted at an intermediate velocity (5.0 ± 0.02 m/s). From CT scans, it was determined that pelves #90 and 92 each fractured at this speed, while pelves #91 and 93 did not. The latter two specimens were impacted again at 6.6 ± 0.2 m/s, resulting in a fracture of pelvis #93. Pelvis #91 still had not fractured and was impacted a third time at 8.3 m/s, at which point it fractured. All resulting pelvic fractures were AIS ≥ 2.

Impact forces were recorded by the 22-kN load cell attached between the impact mass and the striker using DYNATUP™ 930-I software (Cambridge, MA, USA), at a sampling rate of 20.5 kHz. A velocity trigger (Instron Corp., Canton, MA, USA) recorded the impact (maximum) velocity, V_i, of the impact mass. The ratio of the total impact mass (22.1 kg) to the mass behind the load cell (21.6 kg) was determined to be (1.02) and was used to calculate the inertially compensated (Bouquet et al., 1994) impact force, $F(t)$, from which, impactor deceleration, a(t), impactor velocity, $V(t)$, and impact energy, $E(t)$, were calculated. In addition, the impulse, I, was calculated as the area beneath the force-time curve. F_{max} was defined as the maximum force recorded, and E_{peak} was the energy dissipated at the time of F_{max}. For the fracture-producing impacts, F_{max} was taken as the force to fracture. All data was filtered according to SAE J211 (2000).

The two-dimensional displacements of two infrared-reflective markers, one attached to the impact mass and the other attached to the contralateral support of the seat (Figure 4), were captured using the infrared camera system. C_{max}, was calculated as the maximum horizontal change in separation between the two reflective markers following initial contact, divided by the original width of the pelvis at contact. The maximum viscous criterion, $(VC)_{max}$, was determined as the maximum product of the corresponding impactor velocity, $V(t)$, and pelvic compression, $C(t)$.

Linear regression was used to evaluate the association between fracture parameters (F_{max}, I, C_{max}, VC_{max}, and E_{peak}) and the variables BMD and T. Unpaired t-tests were employed to determine significant differences between the results of the no-fracture and fracture-producing impacts. Analysis of covariance (ANCOVA) was performed to compare the resulting equation relating fracture force to BMD against that found for the isolated pelvic bones (females only) in Phase 1. Logistic regression was used to evaluate the association between fracture parameters and the probability of fracture. Generalized Estimating Equations (GEE) were used to produce the logistic regression equations, from which the 25% probability of AIS ≥ 2 was calculated for each parameter. The consistent threshold (Domenico and Nusholtz, 2003) method was also employed to compare against the GEE analysis. These statistical analyses were performed with a level of statistical significance $\alpha \leq 0.05$ (two-sided). As done in Phase 1, fracture parameters were used to indirectly assess whether the current impact conditions were similar to previous full cadaver studies (Cavanaugh et al., 1990; Viano, 1989; Cesari and Ramet, 1982; Nusholtz et al., 1982; Cesari et al., 1980).

RESULTS

Drop tower impacts on isolated pelvic bone-ligament structures

Five of the six unpadded impacts resulted in pelvic fractures (Table 3). Pubic rami fractures were most common, followed by fractures of the sacral wing, iliac wing, and ischium. There was one both-column acetabular fracture. For the six padded impacts, all four fractured pelves experienced medially displaced pubic rami fractures.

Table 3. Impact Response Values In Phase 1.

Pelvis	F_{max} (kN)	E_{peak} (J)	t_{peak} (ms)	Fractures
#43[1]	4.08	73.4	11.3	R. inf/sup rami, ishium, and both-column acetabulum & L. inf/sup rami, iliac wing
#56	5.88	96.1	9.50	L. inf ramus[3] and anterior wall cracks[3] & L.anterior wall cracks[3]
#57	2.59	53.5	12.8	R. inf/sup rami, iliac wing, and sacrum
#58	4.27	64.8	14.3	R. sacral wing and anterior wall crack[3] & L. nf/sup rami and sacral wings
#59	4.99	128.3	19.5	None
#63	2.47	33.7	10.6	R. sup rami, iliac wing, sacral wing, and ischium & L. superior rami
Mean ± S.D.	4.05 ± 1.34	75.0 ± 33.4	13.0 ± 3.58	
#60[2]	4.38	98.6	20.6	None
#61	4.11	147.0	32.3	None
#75	3.52	101.1	26.2	R. inf/sup rami, iliac wing, and sacral wing
#76	2.29	84.9	21.1	R. inf/sup rami, iliac wing, and sacral wing[3]
#77	1.00	36.7	13.7	R. inf/superior ramus, iliac wing, and sacral wing & L. inf/sup[3] rami
#78	2.35	87.8	21.0	None
Mean ± S.D.	2.94 ± 1.29	92.7 ± 35.4	22.5 ± 6.23	

[1]Unpadded group.
[2]Padded group.
[3]Non-displaced fracture.

The motion capture system data was synchronized with the force-time plots. Displaced fractures, shown by abrupt motion of the markers in the camera data, coincided with unloading during the impact event (Figure 5) enabling accurate estimation of fracture forces. Linear regression showed strong correlation (R^2 = 0.89, $p < 0.0002$) between F_{max} and total hip BMD of the nine fractured pelves (Figure 6). Linear regression showed poor correlation between total hip BMD and E_{peak}. The mean total hip BMD for the fractured pelves (0.73 ± 0.28 gm/cm^2) was not significantly lower than that of the unfractured pelves (0.87 ± 0.13 gm/cm^2). Similarly, BMD was not significantly different between the unpadded (0.86 ± 0.23 gm/cm^2) and padded (0.67 ± 0.27 gm/cm^2) impacts. An unpaired t-test showed that the mean t_{peak} for the padded impacts (23 msec) was substantially longer ($p < 0.01$) than for the unpadded impacts (13 msec).

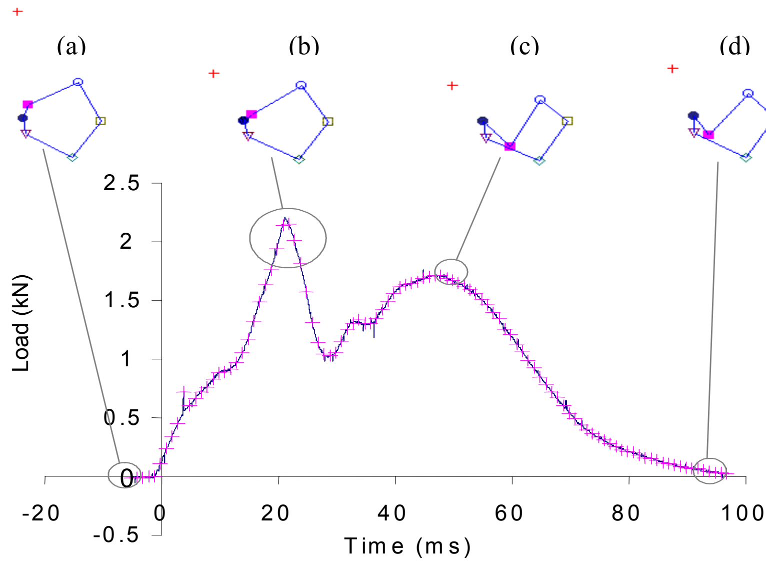

Figure 5: Force-time plot with corresponding camera images (above) for the padded impact of pelvis #76. (a) Velocity trigger initiates data capture prior to impact, (b) fracture of right superior pubic ramus at 25.75 ms, (c) maximum impactor displacement at 44.75 ms, (d) rebounding impactor loses contact with pelvis at 96.75 ms. The single marker on the falling impactor is shown as a cross.

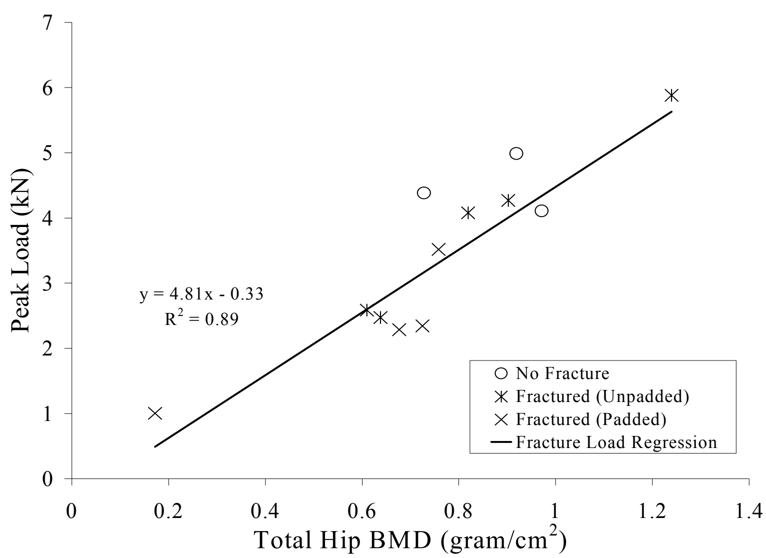

Figure 6: Linear regression demonstrated a significant positive correlation between total hip BMD and the fracture forces in the drop tower impact tests.

Pneumatic impacts of intact pelves

The impacts produced fractures of the pubic rami or pubis in all 10 pelves (Table 4). The next most common fracture was in the sacral ala (n = 9) followed by the posterior iliac crest (n = 3) and acetabulum (n = 2). For the fracture-producing impacts, the results were significantly greater than the no-fracture impacts (p < 0.005) for each parameter with the exception of E_{peak} (p = 0.13).

Linear regression was performed on each fracture parameter as functions of BMD (gm/cm^2), T (cm), and the interaction term (BMD*T). For all parameters, the interaction term was not significant, indicating that BMD and T were independent variables; therefore, the interaction term was removed from subsequent analyses. For F_{max}, both BMD ($p < 0.004$) and T ($p < 0.05$) were significant predictors, resulting in the following equation

$$F_{max} = 6.09*(BMD) + 0.3*(T) - 1.336 \qquad (R^2 = 0.74; p < 0.01). \qquad (3)$$

For all other fracture parameters, BMD and T were not additive predictors of fracture. By means of linear regression, the fracture force was analyzed solely versus BMD for comparison against the regression for female pelves tested in Phase 1 (Figure 7). ANCOVA revealed that for a given BMD the force to fracture the intact pelves was significantly higher than for the isolated pelves ($p < 0.002$). Impulse demonstrated a significant correlation with T but not BMD. Linear regression revealed poor relationships between T and E_{peak} ($p = 0.20$), C_{max} ($p = 0.28$), and VC_{max} ($p = 0.17$) and between BMD and E_{peak} ($p = 0.41$), C_{max} ($p = 0.78$), and VC_{max} ($p = 0.79$).

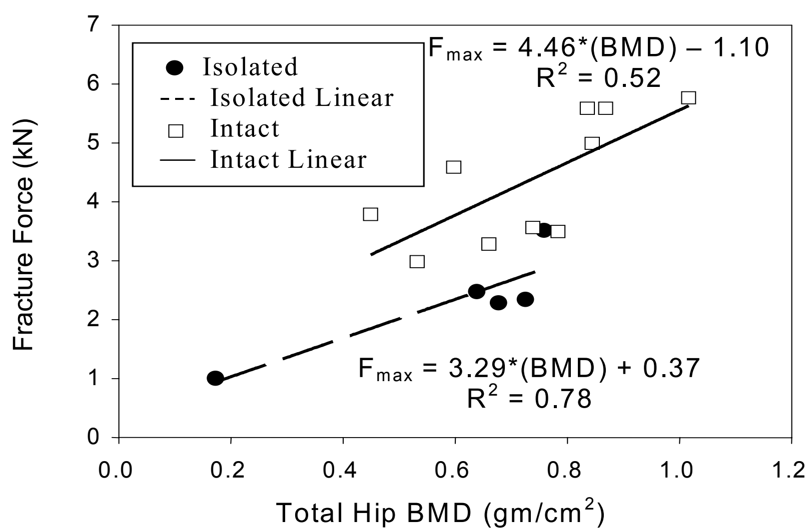

Figure 7: Fracture forces versus BMD for female pelves, both isolated and intact. The linear regressions indicated similar slopes with higher overall values for the intact specimens.

The analysis of the no-fracture and fracture-producing results revealed significant logistic associations with F_{max}, E_{peak}, and I; the associations with C_{max} and $(VC)_{max}$ were borderline significant (Table 5). The GEE tolerances at 25% probability of fracture (AIS ≥ 2) were as follows: F_{max} = 3.16 kN; E_{peak} = 43.85 J; I = 95.61 N-s; C_{max} = 19.77%; and $(VC)_{max}$ = 0.61 m/s. The tolerances found using the CT method were in fairly close agreement with the GEE value with the exception of E_{peak}. At 25% probability of fracture these tolerances were as follows: F_{max} = 3.28 kN; E_{peak} = 26.27 J, I = 89.41 N-s; C_{max} = 24.67%; $(VC)_{max}$ = 0.76 m/s.

Table 4. Impact Responses For The Pneumatic Impacts Of Intact Specimens.

Pelvis #	V_t (m/s)	F_{max} (kN)	C_{max} (%)	$(VC)_{max}$ (m/s)	I (N-s)	E_{peak} (J)	Fractures
83	3.35	2.46	24.66	0.49	66.9	41.5	None
83	6.69	4.59	38.90	1.09	132.2	112.9	R. inf/sup rami and sacral ala L. inf/sup rami and acet
84	3.32	1.83	17.10	0.30	52.2	27.5	None
84	6.28	3.50	32.38	0.91	120.0	122.6	R. inf/sup rami and sacral ala
85	3.39	2.07	20.77	0.34	59.8	26.3	None
85	6.29	4.99	32.19	0.92	114.5	124.7	L. sup rami at pubo-acet junct L. non disp acet and sacral ala
86	3.31	1.63	23.72	0.37	57.5	62.8	None
86	6.27	3.79	37.84	1.14	-----(1)	110.1	L. non disp sup pubis and sacral ala
88	3.28	2.56	20.99	0.31	66.5	40.5	None
88	6.69	5.77	29.55	0.88	123.1	110.1	L. inf/sup rami
89	3.19	1.60	17.66	0.31	52.2	74.5	None
89	6.31	3.28	41.71	1.30	129.0	110.2	L. inf rami at ischial junction, sacral ala and post iliac crest, R. pubis
90	4.99	3.57	31.34	0.76	97.1	49.86	R. inf/sup rami L. inf/sup rami and sacral ala
91	5.03	3.32	-----(2)	0.85	107.8	143.7	None
91	6.39	3.83	42.22	1.33	124.4	196.9	None
91	8.34	5.59	45.89	1.85	167.0	309.6	R. inf/sup rami
92	4.99	2.99	32.41	0.79	89.4	65.2	L. post iliac crest, sacral ala, and inf/sup rami, R. inf rami and pubis
93	4.97	3.55	-----(3)	-----(3)	86.4	70.5	None
93	6.76	5.59	44.46	1.33	123.5	116.0	L. inf/sup rami, and post iliac crest symphyseal disruption, R. sacral ala

[1]No impulse calculated due to loss of signal from load cell after peak force occurred.
[2]No C_{max} found due to loss of signal from camera during impact; however, enough camera data was recorded to obtain VC_{max}.
[3]No C_{max} or $(VC)_{max}$ found due to loss of signal from camera during impact

Table 5. Logistic Regression Results Generated By The GEE Method For All Impacts.

Logistic Result	F_{max} (kN)	C_{max} (%)	$(VC)_{max}$ (m/s)	E_{peak} (J)	I (N-s)
p	0.0035	0.0553	0.0612	0.0020	0.0022
Tolerance (1)	3.16	19.77	0.61	43.85	95.61

[1]Fracture tolerances represent 25% probability of pelvic fracture (AIS ≥ 2).

DISCUSSION

Drop tower tests

The results of phase 1 showed that total hip BMD correlated with fracture loads in our drop tower impacts of isolated human pelves. This was the first evidence that BMD may prove useful in predicting pelvic fracture risk in automotive side impacts. The average peak load (3.49 ± 1.38 kN) were close to those measured in similar drop tower impacts of isolated pelves (Guillemot et al., 1998), but were lower than those reported in sled and pendulum tests on full cadavers (Zhu et al., 1993; Cavanaugh et al., 1990; Viano, 1989; Cesari and Ramet, 1982; Nusholtz et al., 1982), as listed in Table 6. Unlike sled and pendulum devices, which permit the specimen to move in response to impact forces, specimens in Phase 1 of our study were rigidly constrained on the contralateral side and the falling mass of our impactor continued to load the specimen after the initial impact event. Comparisons with previous studies, therefore, should be mindful of different test conditions. Nevertheless, the times to peak load in the padded (t_{peak} = 22.5 ± 6.23 msec) and unpadded (t_{peak} = 11.7 ± 6.2 msec) tests as well as the predominance of rami fractures (LC-I) seen in the present impacts are consistent with previous lateral impact studies of full body cadavers (Zhu et al., 1993; Cavanaugh et al., 1990; Viano, 1989; Cesari and Ramet, 1982; Nusholtz et al., 1982). In many of the pelves, the posterior iliac wing and anterior sacrum fractured in addition to the rami, which is typical of nearside motor vehicle crashes (Gokcen et al., 1994).

Table 6. Pelvic Response Comparisons With Previous Impact Studies.

Study	Impact Method	Impact Condition	Mean Peak Load (kN)	Mean Fracture Load (kN)
Cesari and Ramet (1982)	Cannon: full cadaver	Unpadded Padded	8.38 ± 2.96 (n = 55) 8.81 ± 1.73 (n = 4)	8.55 ± 3.25 (n = 16) 9.29 ± 1.76 (n = 3)
Nusholtz et al. (1982)	Pendulum: full cadaver	Unpadded Padded	7.09 ± 1.96 (n = 7) 11.0 ± 4.55 (n = 5)	6.54 ± 2.04 (n = 5) 13.0 (n = 1)
Viano (1989)	Pendulum: full cadaver	Unpadded	8.11 ± 3.18 (n = 14)	9.78 ± 0.52 (n = 2)
Cavanaugh et al. (1990)	Sled: full cadaver	Unpadded Padded	9.55 ± 3.17 (n = 8) 5.59 ± 0.99 (n = 4)	11.2 ± 2.83 (n = 5) 6.41 (n = 1)
Zhu et al. (1993)	Sled: full cadaver	Unpadded Padded	9.97 ± 3.36 (n = 8) 5.67 ± 1.12 (n = 9)	11.7 ± 3.04 (n = 5) 6.45 ± 0.41 (n = 2)
Guillemot et al. (1998)	Drop tower: isolated pelvis	Unpadded	2.78 ± 1.22 (n = 12)	2.37 ± 0.76 (n = 10)
Present Study: Phase 1	Drop tower: isolated pelvis	Unpadded Padded	4.05 ± 1.34 (n = 6) 2.94 ± 1.29 (n = 6)	3.86 ± 1.40 (n = 5) 2.29 ± 1.03 (n = 4)

The primary advantage associated with removing the soft tissues was the ability to monitor the deformation and fractures of the pelvic ring. Soft tissue removal also eliminated differences in the amount of muscle and adipose tissues surrounding the greater trochanter, a potential source of variability in pelvic impacts (Nusholtz et al., 1982). The important disadvantage was that by removing soft tissues and abdominal contents, their contributions to the structural integrity of the pelvis were eliminated. Confirmation of this correlation of BMD with fracture parameters for intact cadaver pelves was needed to support the clinical relevance of these findings, and led to Phase 2 of the present study. The average cadaver age in Phase 1 was approximately thirty years greater than that of victims of pelvic fracture in crash studies (Dakin et al., 1999; Lewis et al., 1996; Gokcen et al., 1994); therefore, further testing is needed to determine if the correlation extends to younger sample populations.

Pneumatic impacts

In Phase 2, the force to fracture the female pelves correlated positively with both BMD and T, as hypothesized [Eq. (1)]. The resulting equation was a better predictor of fracture force than those containing BMD or T alone. We observed that for a given BMD, the presence of soft tissues increased the force required to fracture the pelvis in lateral impacts as compared to the Phase 1 tests on the isolated bony pelves. This is consistent with clinical observations, which showed that subcutaneous fat depth cushioned lateral impacts, lowering the AIS scores for the pelvis and abdomen (Wang et al., 2003; Arbabi et al., 2003).

Equation (1) may be used to predict fracture forces for the independent variables, BMD and T. For example, inserting the average BMD value for a normal 75-year old female (0.728 g/cm^2) (Looker et al., 1998) and the average T from our study (4.13 cm), the equation predicted a 4.34 kN fracture force, which is consistent with the previous female tolerance by Cesari and Ramet (1982). Similarly, incorporating the BMD value for an osteoporotic woman (0.637 g/cm^2) (Looker et al., 1998) and the average T, less one standard deviation, (2.25 cm) resulted in a 3.22 kN fracture force, which is in close agreement with our reported tolerances of 3.16 kN (logist analysis) and 3.28 kN (CT method) at 25% probability of fracture. These are substantially below the 4.4 kN and 5.77 kN tolerances reported previously from trochanteric loading of full cadavers (Cavanaugh et al., 1993; Cesari and Ramet, 1982). Our results imply that a 3 kN force tolerance for pelvic fracture force may be an appropriate value for older women with an increased risk of fracture due to low bone quality and general frailty.

Increases in soft tissue thickness at the greater trochanter attenuated the impacts resulting in increased forces to fracture and impulses. BMD, on the other hand, did not correlate with impulse. Our overall impulse tolerance of 95.6 N-s was in close agreement with the 100 N-s tolerance reported by Cesari et al. (1980) using full cadavers. E_{peak}, C_{max}, and $(VC)_{max}$ were observed to be independent of both BMD and T. Increased trochanteric tissue thickness also increased the energy dissipated at fracture; however the relationship was not significant. The compression-based fracture tolerances found using logistic regression appear conservative when compared to previous pendulum and sled impacts of whole cadavers. The pelvic compression tolerance of 19.77% is less than the 27.4% reported by Viano (1989) and 32.6% observed by Cavanaugh et al. (1990). Furthermore, our $(VC)_{max}$ tolerance of 0.61 m/s is roughly 43% less than the 1.07 m/s determined from whole cadaver sled tests (Zhu et al., 1993). While Viano (1989) measured full pelvic compression, Cavanaugh et al. (1990) and Zhu et al. (1993) measured compression of the struck half of the pelvis only, which would tend to produce higher measures of compression and viscous criteria.

Potential limitations to Phase 2 of the present study include the use of our pelvis-femur sections, which required the addition of mass to approximate the inertial effects from missing upper and lower extremity body segments. The contact area over which the load was applied in the present study (77.4 cm^2) concentrated almost entirely over the greater trochanter. Trochanteric loading has been shown to produce lower fracture values than loading both the greater trochanter and iliac wing (Haffner et al., 1985). The time to peak load in the present study was approximately 10 ms and the duration of the impacts was approximately 20-50 ms, however, which are consistent with times reported in previous testing of torsos and intact cadavers (Bouquet et al., 1994; Viano, 1989; Nusholtz et al., 1982). In addition, we recognize that our data is censored (i.e. the actual fracture force may not correspond with the maximum force recorded, F_{max}), hence the actual fracture forces may actually be slightly lower than the reported values.

The use of multiple impacts to each specimen in Phase 2 permitted us to obtain additional data points for logistic analyses and tolerance calculations, as done by others (Viano, 1989; Cesari et al., 1982). It also created the possibility for micro-fractures that may not have been visible in our radiographic examinations. Such micro-fractures may have affected the impact response of the pelves under subsequent impacts by reducing peak forces and increasing compression levels. The fracture types seen in the present study (e.g. rami and acetabulum) were consistent, however, with those typically observed in experimental side impacts (Beason et al., 2003; Cavanaugh et al., 1990; Viano, 1989; Cesari and Ramet, 1982; Cesari et al., 1980) and in motor vehicle crashes (Dakin et al., 1999; Guillemot et al., 1997; Gokcen et al., 1994; Seigel et al., 1993).

CONCLUSIONS

In conclusion, the present results indicate that bone mineral density and trochanteric soft tissue thickness affect pelvic fracture in experimental impacts to the greater trochanter. The primary conclusions may be drawn:

1. BMD correlates with the force to fracture the pelvis with or without the presence of soft tissues;
2. The force to fracture was significantly higher for intact pelves than for the isolated pelvic bone-ligament specimens;
3. The tolerance of the intact female pelves to fracture was just over 3kN, which is lower than previously reported tolerances for the 5^{th} percentile female.

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DISCUSSION

PAPER: **Bone Density and Trochanteric Tissue Thickness Affect Fracture of the Female Pelvis in Side Impact Experiments**

PRESENTER: ***Brandon Etheridge, Department of Biomedical Engineering, University of Alabama at Birmingham***

QUESTION: *Erik Takhounts, NHTSA*

I have a couple questions about your test set-up. First question: As a boundary condition, you have a constraint on the other end?

ANSWER: Yes.

Q: How does that affect your result?

A: Well, I can't say how it affect our results because we didn't do it without it there, but we looked at it as possibly—I mean, you could say it's similar to having a console in a car. You know, I can't tell you what would happen without it there. I just know what happened with it there.

Q: Okay, when you measure your force with a load cell—

A: Yes.

Q: The load cell is accelerated. Do you compensate, initially, for this acceleration of the load cell? Do you have an accelerometer there?

A: We did put an accelerometer on our impacting mass, but we saw that it had waves running through it and it looked like it was vibrating in its resonance and it was getting extremely large peak loads. It was, did not look trustworthy at all and we tried to filter it out. We even spoke to the manufacturers of the machine and they did—They couldn't even tell us what was going on.

Q: Okay. So this load cell data, this peak force: That actually includes the actual acceleration of the load cell itself.

A: Yes, but we set up the cadaver at the seat to where it strikes immediately once the piston has extended itself fully, so we like to think that the acceleration is constant at the time of impact.

Q: So…

A: I don't know.

Q: Okay. Thank you.

A: Thank you.

QUESTION: *Guy Nusholtz, DaimlerChrysler*

Just a question to follow up on Erik's question if you could remember this data: When you looked at the load trace, did you see a small ripple similar to what you saw on the acceleration?

ANSWER**:** No.

Q: No, you didn't. Okay. So it wasn't a fact of just the scaling for compensation. I mean, it solves Erik's problem if you have the small ripple because then that implies that you've got a very small load that's due to the acceleration of the mass. Otherwise, you don't know what the additional force is associated with the acceleration of the mass in front of the load, associated with the mass of the load cell. But if you didn't see that—

A: No, we did not see that.

Q: Typically in a lot of load cells, I see both accelerations, but the loads are very small. You scale the acceleration to get to zero it out and then you're, you have a compensated load cell. You impacted the subjects multiple times?

A: Yes.

Q: Until it fractures? How do you know that you didn't weaken the material?

A: They were—We don't know we didn't weaken it. They were radiographed in between to check for fractures, but we, you know, other people have done that and there was—We needed to increase our n. We wanted no fracture cases and fractured cases. The load—The first impact was at a pretty low speed that we were pretty confident wouldn't, you know, wasn't a destructive impact. But at the same time, we can't be sure we didn't weaken it.

Q: Okay. What is the reason for the soft tissue, the increased tissue? You basically created a plane and you rise up as you increase the amount of soft tissue in that one graph, if I read it correctly. That one. Yeah. Soft Tissue Thickness. Why does the increase in soft tissue increase the fracture force?

A: Well, it's kind of what Dr. Wang was talking about earlier: It absorbs. It's an energy absorbing material and it requires a large force to, when that energy is absorbed, to actually break the bone inside.

Q: But, the amount of force is going to be related to the mass of the soft tissue between the—You know, there's no force except for the mass times acceleration of the soft tissue. So you'd be arguing that the force reduction ends up being related to the mass between the two, the mass that's accelerated. You should be able to go through and do a hand calculation to see if that's real or not.

A: I have to look at that.

Q: I mean, if you take a spring—

A: Sure.

Q: And you put a force on it, even though there's energy absorbed as you come along, the force at Point 1 and Point 2 does not change. But if you have mass in there, then the force'll reduce. It could also possibly be related to the way you're going to distribute the load on the acetabulum. And so the distributional load coming through may be as important as the actual load. So you have other complexities associated with it. Okay. Thanks.

A: Thank you.

QUESTION: *Srini Sunderarajan, Ford Motor Corp.*
Couple of questions. One is: You did, I think, 10 or 12 cadavers tests.

ANSWER: Ten.

Q: Did you see a different fracture pattern depending upon the amount of soft tissue that was there?

A: No.

Q: So, what type of fractures did you see?

A: Pubic rami were the predominant. We saw a few sacral wing and a couple of acetabular. Almost—I think, actually, all of them had a pubic rami fracture.

Q: So you're suggesting that even though there's thick tissue, the load distribution happens to be similar in all the cases?

A: Yes. The primary case—The primary fracture, the driving force is, especially when you're doing greater trochanteric impacts, is the bending of the pelvis in the rami or the small, the weak link of the pelvis, and the bending force is usually what breaks that regardless of whether the tissue's there or not.

Q: The second question is: I could not understand the advantage of isolating a pelvis with all the tissue and doing a testing versus a full cadaver testing. Do you have any particular advantage--?

A: Why did we take the soft tissue off?

Q: No, no. I'm saying: In the second series of tests, you took an isolated pelvis with all the tissue and ran the test. Now, what is the advantage of that over running a full cadaver test?

A: We can't afford to buy a full cadaver. We didn't—I mean, they're given to us sectioned. We didn't just cut it out.

Q: Okay. Thanks.