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Assessment of g-forces on the Crazy Frogs amusement ride

ERG/04/20

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EXECUTIVE SUMMARY

OBJECTIVES

- To record the g-forces / accelerations that can be experienced by passengers on the Crazy Frogs ride operated by Mr C Horne.

- To determine the risk of vertebral injury due to the compression forces resulting from the ride g-forces / accelerations and the ride motion patterns.

- To provide recommendations on steps that should be taken to control the risk of injury from exposure to ride g-forces / accelerations and the resulting compression forces

MAIN FINDINGS

- The peak ride g-forces / accelerations were found to be highest during the bouncing motion (up to 4.1g peaks).

- If there were no vertebral strength reducing effects due to repetitive compressions, the IP's L1 vertebra ultimate compression strength (UCS) would need to have been in approximately the weakest 1% of the general adult population (1354N or less) for the injury to occur during the ride motions that were recorded.

- There is evidence that after 10 repeated cycles, a compression force equivalent to just 60% to 70% UCS results in a 9% probability of the vertebrae fracturing.

- When a 60% effective L1 UCS is assumed, the IP's estimated full (100%) L1 UCS is within the ranges of the available data on L1 UCS in the general population.

- The risk of vertebral compression injury to any particular individual is determined by the ratio of their L1 UCS to their weight. People may be grouped into 'weight populations' within which there are different ranges of L1 UCS, with heavier people tending to have higher L1 UCS. Based on the data available it is not possible to identify the probability of a specific / high-risk weight / L1 UCS discrepancy occurring.

- An effective ride acceleration / g-force safety margin may be calculated based on assumptions about the greatest weight / L1 UCS discrepancy that is considered feasible, For example an assumption that a heavy male (e.g. 95%ile weight) may have a 1%ile L1 UCS would lead to maximum acceptable z-axis g-forces of 2.2g (or 3g if repetition effects are not considered).

- The highest compression force estimated to act on the IP's L1 vertebra from the bouncing motion is 23N higher than the lowest recorded L1 USC in the scientific data.

- The peak z-axis g-forces are generally higher than the majority of other rides which HSL has recorded data from. With one exception, any higher g-forces on the other rides assessed by HSL occur as single / infrequent events and not repeated bounces.

- Bone Mineral Density and vertebral endplate cross-sectional area are key factors in determining L1 UCS. These factors are themselves influenced by age, gender, physique / build, illness etc. There is no indication that a person would be aware of having a sufficiently low L1 UCS that their body weight may be high enough to cause a sufficiently high discrepancy that could cause a risk of compression fracture.

- Spine flexion increases the risk of vertebrae fracture under axial compression. Factors which could lead to spine flexion on the Crazy Frogs ride are the relatively quick onset of acceleration, fatigue of posture support muscles and running the ride in reverse.

- The main groups considered to be at increased risk of injury are:

  - Older people;
  - Females;
  - Taller people (heavy but with narrow bone structure / low BMI);
  - People with uneven / top-heavy weight distributions;
  - People with bone mineral deficiency conditions such as osteoporosis;
  - People suffering from other spinal / bone disease (tumors etc.);
  - People with conditions which reduce body weight / nutrient metabolism.

RECOMMENDATIONS

- Bounce motions should be avoided if they include peak g-forces / accelerations at the levels that were recorded (up to 4.1g in z-axis).

- If bouncing continues to be used, peak g-forces / accelerations should be limited to a maximum value based on a greater margin of safety than currently exists.

- All ride motions / accelerations should be assessed to determine the appropriate margin of safety.

- Bounces, if used, should occur for shorter periods (i.e. there should be fewer consecutive bounces) and should not occur while the ride is being rotated in reverse.

- Effective signs, instructions and operator awareness should all be used to reduce the likelihood of a higher-risk individual using the ride (however, this should only be considered as a back-up to the increased safety margin described above).

- Restraints which ensure passengers do not lift off their seats and do not adopt spine flexion postures are likely to further reduce any risk of vertebral compression injury (once again, these should only be considered as a back-up intervention).

1.0 INTRODUCTION

This is an updated version of a report which examined the g-forces on the 'Jumpin' (Jumping Frogs) ride (HSL Report ERG/04/02). This investigation of the g-forces is due to a member of the public (IP) sustaining an injury whilst on the Jumpin' ride, a Crazy Frogs ride operated by Mr C Horne.

It was initially understood that the IP had suffered a compression fracture to vertebra Nr. 17 (or the Thoracic - T10 vertebra). However, it has since become known that the IP suffered the fracture to her L1 vertebra (the first lumbar vertebra). The injury was specifically diagnosed as a Burst compression fracture. The site of the IP's Burst fracture (L1) is typical of this type of injury, which almost always occurs in the thoracolumbar junction area of the spine (i.e. around the 12th Thoracic vertebra and the 1st Lumbar vertebra).

Burst fractures are specifically characterised by the loss of height of the entire vertebral body and the margins of the vertebral body are spread out in all directions. They are also characterised by a fragment of the vertebral body being retropulsed into the spinal canal, which can compromise / bruise the spinal cord causing neurological problems below the injured vertebra. Burst fractures are caused when a sufficiently high axial compression force acts on a vertebra; they typically result from high impact events such as jumping from height, a heavy downwards loading on the shoulder region, or in vehicle accidents (Willen et al, 1990).

Following the incident, I was asked by Dr Stuart Arnold (HSE Inspector) to measure the g-forces (accelerations) on the ride to determine whether they are excessive and represent a significant hazard in terms of similar spinal compression injuries.

1.1 Aim

This investigation has three aims:

- Record the g-forces which passengers may be exposed to on the ride;
- Determine the likelihood that injury (in particular spinal injury) may be caused by the ride forces;
- Provide recommendations on ways to reduce risks to passengers.

1.2 Visits

Two visits were made in order to measure the ride g-forces. On the first visit (11 August 2003 at St Andrews street fair) I measured forces on the Jumping Frogs ride using an accelerometer fixed to the outer seat of a passenger car. Following this visit I had two main concerns:

- A review of the video recording of the measurement runs suggested that during testing the ride did not appear to be driven as fast / vigorously as Crazy Frogs rides I had seen in operation at another fair;

- The accelerometer which I fixed to the seat measured the motions / accelerations of the ride but may not have recorded the forces actually experienced by the passengers. The data from the initial tests and subjective impressions from people who had used the ride suggested that at the top of an upwards motion and the start of the downwards motion there may be a temporary loss of contact with the seat. If this were so it would potentially mean an impact force at the bottom of the ride movement as the ride slows and begins moving upwards again (a batting effect). This impact force may result in higher g-forces than would be measured by an accelerometer fixed to the seat.

For these reasons and because HSE learned that the ride was being driven in reverse at the time of the incident, a second visit was made to record ride forces under conditions which were more representative of the ride at the time of the incident.

On the second visit (9 December 2003 at Horne's Yard in Glasgow) I attached the tri-axial accelerometer (which I use to measure ride g-forces / accelerations) to an anthropometric dummy weighing approximately 70kg. Figure 1 shows the location of the accelerometer. This attachment point was considered appropriate because of its proximity to the thoracolumbar junction region of the spine where the injury occurred. A dummy was used to simulate a passive passenger (i.e. fixed / pivoting around the hand but not braced against the backrest) so that any impact forces could be recorded.

Figure 1. Location of accelerometer on dummy (arrow indicates accelerometer location under rib-cage)

This report focuses primarily on the g-forces recorded on the second visit (9th December) because they are considered to be more representative of the levels of acceleration which passengers may typically experience.

1.3 Test Conditions

The speed and acceleration of the ride arms / passenger movements depends on the level of compression driving them. The compression levels on the Crazy Frogs ride are dependent on the operator judging how great the load is on the ride. If there is a high load (a full load containing numerous adults) the operator will use a relatively high compression level. If there is a low load (e.g. a few children) the operator will use a lower compression level. For a specific weight, higher compression levels will result in higher g-forces and higher rates of change of g (jerk levels).

During testing on the 9th December, HSE requested the ride to be run at the highest compression level which would not cause damage to unoccupied cars / arms. This was approximately 6.6kgf/cm² (approximately 94 psi) although the levels drop slightly throughout a full ride cycle. The units on the compression readout in the operator's control booth are kgf/cm².

2.0 ANALYSIS OF G-FORCES

G-force data was measured at 40Hz using an Entran® +/-25g accelerometer. The data was recorded onto a data logger (HSL proprietary equipment: 'efairlog') and downloaded onto a laptop computer using HSL proprietary software. To analyse the g-force data I took the following steps:

1. Performed a low-pass software filter (10Hz) to remove noise.

2. Imported data file into Excel and identified any z-axis data points that exceeded 2g and –0.1g

3. Identified peak values and match these with types of ride motion. From the initial tests at St Andrews, 2 key types of ride motion were identified. These types of motion, described below, can be alternated between cars to create different overall patterns of motion:

    a. Low-frequency high-amplitude 'wave' motion;

    b. High-frequency low-amplitude 'bouncing' motion.

4. Calculated the typical jerk (rate of onset / change of g) values for periods of the main motion types.

3.0 TEST 1: RESULTS

On test 1 (09.12.2003) the dummy slipped down in the seat so the baseline z-axis g-force was reduced to approximately 0.78g. From the appearance of the overall test graph it is likely that the dummy began slipping down at some time during the first series of bounces. This report therefore concentrates on the g-force data gathered in test 2 (09.12.2003).

4.0 TEST 2: RESULTS

4.1 Peak z-axis g-forces during low-frequency wave movements

Table 1 provides summary data for a series of 12 low-frequency wave movements (the series is considered to be typical of this type of motion).

Table 1. Summary of z-axis g-forces during 12 wave movements.

	Upwards force (at bottom of bounce motion)	Downwards force (at top of bounce motion)
Maximum peak z-axis g-force (g)	3.25	0.09
Minimum peak z-axis g-force (g)	2.81	0.3
Mean peak z-axis g-force (g) (across 22 bounces)	3.16	0.14
Std. Dev. Of peak z-axis g-forces (g)	0.14	0.06
Bounce frequency	0.6 Hz	

Figure 2 shows this set of wave movements.

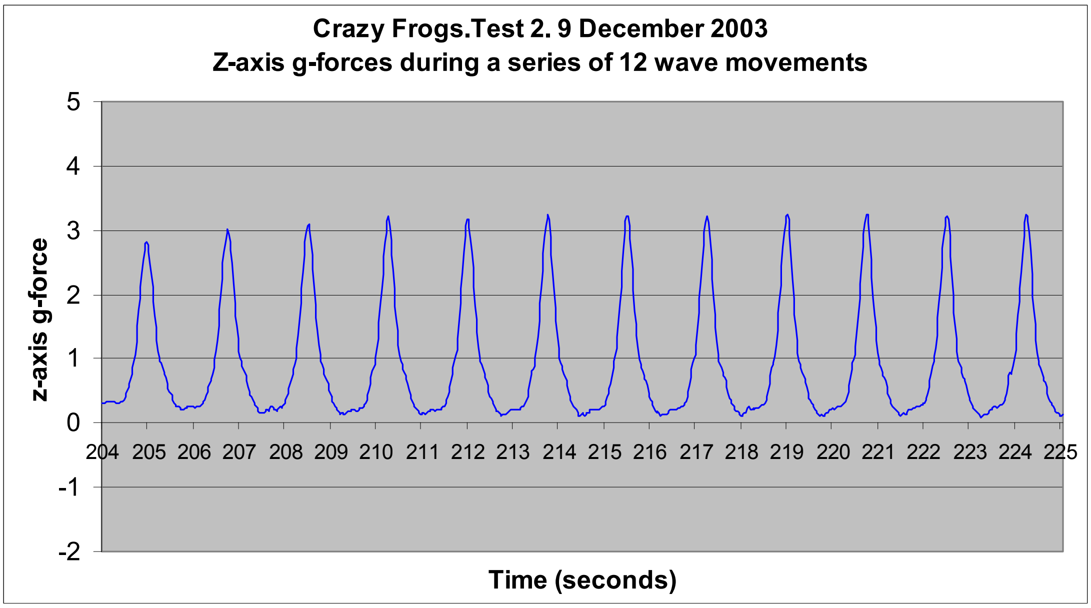

Figure 2. A typical series of low-frequency / high-amplitude wave motions

4.2 Peak g-forces during high-frequency bounces

Table 2 shows a summary of the g-forces measured on a set of 22 bounces during test 2.

Table 2. Summary of z-axis g-forces during 22 bounce motions.

	Upwards force (at bottom of bounce motion)	Downwards force (at top of bounce motion)
Maximum peak g-force (g)	4.11	-1.69
Minimum peak g-force (g)	2.57	-0.03
Mean peak g-force (g) (across 22 bounces)	3.37	-0.69
Std. Dev. Of peak g-forces (g)	0.39	
Bounce frequency	1.2 Hz	

Figure 3 shows this set of bounces.

The peak forces during these bounces were similar to the peak force recorded during the first visit to record ride forces (St Andrews). During the visit to St Andrews I requested Mr Horne to operate the ride in the same way that he would during normal conditions.

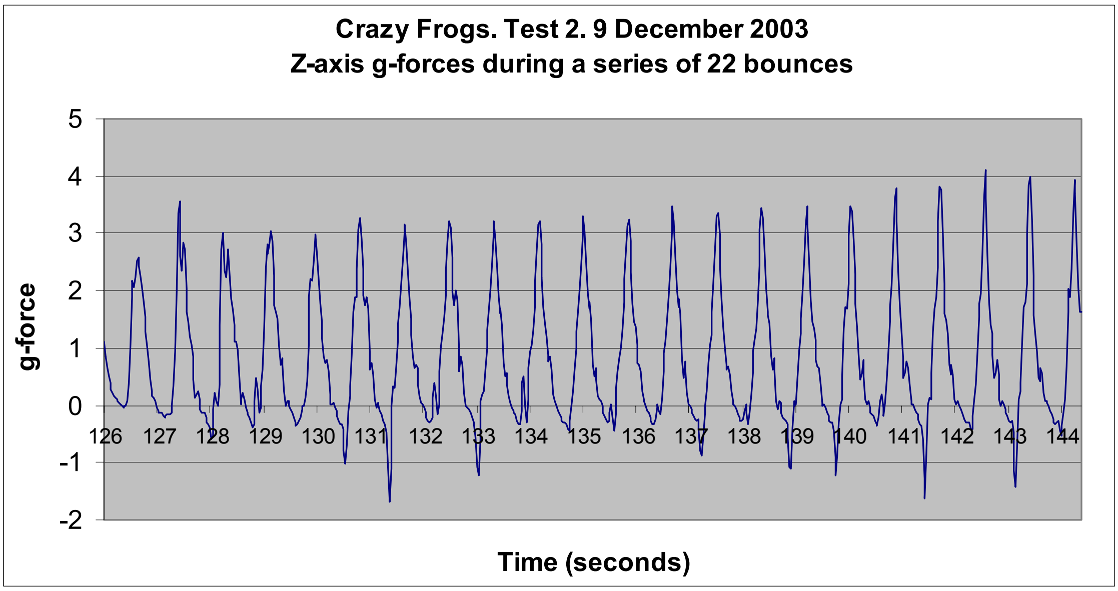

Figure 3. G-forces during a set of 22 bounce motions

5.0 ASSESSMENT OF G-FORCES

5.1 Overview of g-forces

Due to the way the ride is operated, it was not possible to exactly duplicate the motions and forces which may have resulted in injury (the same balance between passenger weight and air pressure could not be exactly replicated). However, during the tests on the 9th December, Mr Horne was asked to operate the ride using the highest air-pressure levels that he felt could be used without causing damage to the ride. The g-forces recorded are therefore considered to be representative of the forces that would generally be experienced by a member of the public. Positive z-axis g-forces on the Crazy Frogs ride are generated by the downwards deceleration and upwards acceleration which occur at the bottom of an up / down motion. The effects of the deceleration and acceleration are the same in terms of the resultant forces (including spinal compression forces) experienced by a passenger. Therefore, unless stated otherwise, in the remainder of this report they are referred to collectively as the rides' accelerations.

The g-force data shows that the highest g-forces on passengers will occur during the 'bouncing' motion (the high-frequency low-amplitude motions). If they are considered as isolated instances, the peak upward forces that were recorded during these movements (generally between 3 and 4g after filtering) are unlikely to lead to injury. The g-force levels we recorded are almost entirely within the +4g to –1.5g TUV advised lower levels for safety (RWTUV Fairground Rides Attractions with Calculated Safety: The strain on passengers, Limit values for Roller Coasters). The g-forces are also within the +5.5g to –2.5g AS3533.1 (1997) recommended maximum ranges for z-axis accelerations. It should also be noted that these recommended limits are referred to the cardiovascular effects of prolonged (>6s) exposure to acceleration e.g. visual blurring and loss of consciousness rather than the compression loading of the spine. The short durations of the maximum accelerations on the Crazy Frogs ride means that there is likely to be very little risk of passengers suffering from these types of cardio-vascular symptoms.

These g-forces are within the range predicted by the manufacturer's report (IM/108/00). In their report two load cases are given as examples. In load cases 1 and 2 the radial accelerations of the car are given as 11.407 rad s^{-2} and 6.75 rad s^{-2} respectively. On the basis of a 4m distance between the arms' pivot point and the passenger seating area, these figures represent g-forces of approximately 4.65g and 2.75g respectively, acting on passengers in outer seats.

5.2 Comparison of g-forces with scientific data on vertebral compression

To establish the level of risk associated with the types of motion / forces on the Crazy Frogs ride, the range of compression strengths of L1 vertebrae need to be considered and the potential cumulative / fatigue effects of repeated compression cycles also need taking into account.

5.2.1 Approximate forces on IP’s L1 vertebra

In order to compare the forces acting on the IP’s L1 vertebra with the levels of force which have been identified by scientific studies as causing compression fractures and specifically burst fractures, it is necessary to determine the potential L1 compression forces caused by the ride’s action. A study by Ruff (1950) calculated that the L1 vertebra supports approximately 50% of body weight. The exact proportion will vary between people depending on the way their weight is distributed however without specific information on the IP’s weight distribution the figure of 50% weight supported by L1 vertebra is considered to be a reasonable estimate.

The IP has informed HSE that at the time of the incident, her weight was approximately 73.9kg, which corresponds to 36.95kg supported by the L1 vertebra.

To calculate the forces acting on the L1 vertebra, the vector of the z-axis acceleration (acting perpendicular to the seat) and y-axis accelerations are used. This approach assumes that the centrifugal force from the rotation of the ride will cause passengers to lean inwards slightly towards the centre of the ride to reduce uncomfortable lateral shear forces on their head / neck. If passengers lean inwards in this way, it is likely they will experience the z and y-axis vector force axially through their spine. The compression forces acting on L1 resulting from the ride’s accelerations have been calculated by multiplying these vector accelerations (ms^{-2}) with the mass supported by the L1 vertebra (kg). Figure 4. shows the L1 compression forces which, for a person of the IP’s weight, are estimated to result from the sequence of high-frequency bounces. Table 3 contains a statistical summary of the final 21 consecutive bounces and the final 10 consecutive bounces in this sequence.

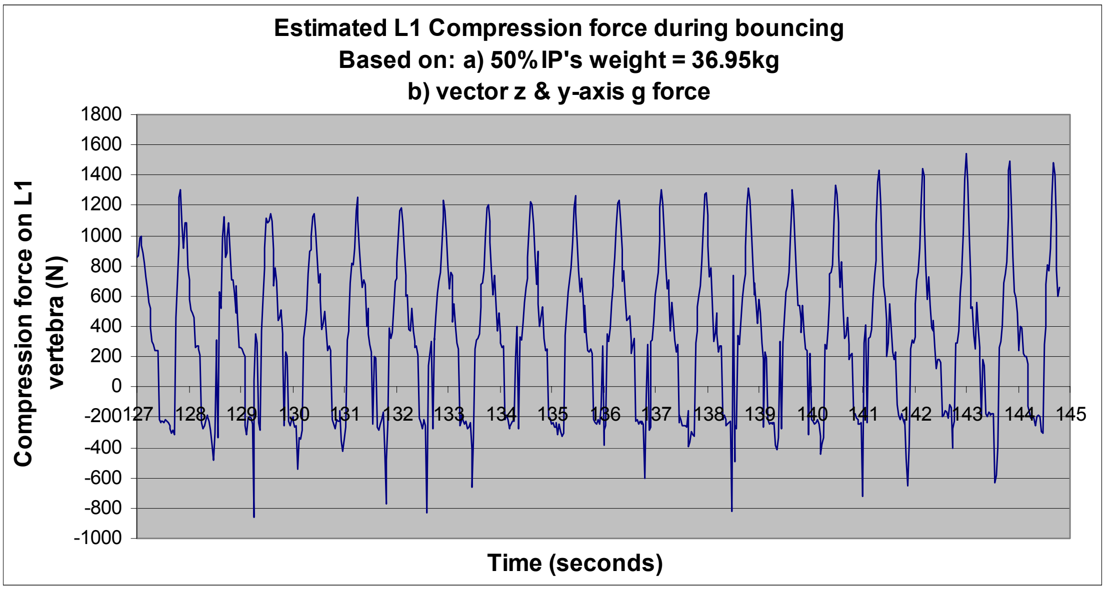

Figure 4. Estimated forces which would act on IP’s L1 vertebra during recorded accelerations.

Table 3. Summary of estimated L1 peak compression force data.

	Data	Peak forces (N)	Mean during highest 50msec of force (N)
21 consecutive bounces to end of graph (Fig. 6)	**Mean *(& Std. Dev)***	1296 *(120)*	1276 *(106)*
	Max	1543	1461
	Min	1125	1116
10 consecutive bounces to end of graph (Fig. 6)	**Mean(& Std. Dev)**	1391 *(95)*	1355 *(86)*
	Max	1543	1461
	Min	1284	1259

The average / range of the peak 50msec of force during each bounce was calculated because in a previous study testing the compressive strength of vertebrae, this was the time period over which forces were applied and caused fracture (Lin et al, no date)

Figure 4 and Table 3 show that during the final 10 bounces of the sequence the forces are higher than at the start of the sequence.

5.2.2 L1 compression strength range

A range of studies have been carried out to determine the compression forces at which vertebra (including L1) will fail. Various terms are used in these studies, for example, compressive strength, breaking load and failure load. What all the force values have in common is that they refer to a level of force which was causes vertebra to fracture and be damaged beyond repair. Certain studies have tested the forces necessary to create burst fractures whilst others have looked at the forces needed to create less catastrophic failures (e.g. stellate fractures on the surface of the vertebral body, wedge fractures etc.). In cases where less catastrophic fractures have occurred, this was typically because the experiment was stopped before more damage was caused to the vertebra. However, it is expected that if further compression force / additional compression cycles occurred to a vertebra which is already weakened by fractures this may result in more catastrophic failures such as burst fractures. For the remainder of this report the loads at which fracture / failure occurs will be collectively termed as Ultimate Compressive Strength (UCS).

In the following analysis the L1 vertebra UCS data is categorised as follows:

- **Burst Fracture Data**

Forces resulting in L1 vertebra Burst fractures.

- **General Fracture Data**

Forces leading to any L1 vertebra fractures (including burst fractures). This data group is split into two as follows:

  - Data from all subjects / studies (excluding data for subjects aged >60 yrs from a study by Hansson et al, (1980);
  - Data from <u>all</u> subjects in Hansson (1980) study (including >60yrs. data).

Appendix 1 contains details of all the L1 vertebra UCS data which was used to compare with the estimated compression forces acting on the IP's L1 vertebra due to the accelerations during bouncing. Table 4 below contains a summary of the UCS data.

Table 4. Summary of L1 vertebra UCS data groupings.

L1 UCS Data Group	**Nr. of subjects**	**Min L1 vertebra UCS (N)**	**Max L1 vertebra UCS (N)**	**Mean L1 vertebra UCS (N) (& Std. Dev.)**
Burst only	26	2542	12535	6874 *(2350)*
All types of fracture (<=60yrs only from Hansson, 1980)	73	2030	12535	5911 *(2464)*
All types of fracture – all data used (incl. >60yrs from Hansson, 1980)	90	1520	12535	5299 (2586)

It should be noted that all of the L1 UCS data has come from adult vertebra. The study by Hansson et al (1980) allows the data for subjects aged >60yrs to be filtered out. The data from other studies which are known to contain some L1 UCS from subjects aged >60 such as Shono et al (1994) and Shirado et al (1992) cannot be similarly filtered because specific ages are not linked with the L1 UCS data. Experiments to measure vertebral compression strengths use vertebra taken from cadavers. This methodology somewhat skews the age of the subjects towards older adults, however it should be noted that none of the studies that I refer to tested vertebra which came from people with known bone mineral deficiencies / abnormalities prior to testing.

Much of the burst data and general data does come from adults aged 30 and upwards, i.e. similar in age to adults who might accompany children on rides at fairs / amusement parks. Therefore any further reference to population or 'general population' in this report refers primarily to the adult population aged approximately 30 and older.

The Burst fracture data show a substantially higher mean force / strength and a higher minimum recorded value for force leading to this type of fracture. This may be because when a single compression event is used in experiments to cause a fracture, a greater force might be necessary to cause the vertebra to burst, than would be needed to create less severe fractures / permanent deformations etc.

5.2.3 Effects of repetition on compression forces required for vertebral fractures

Research has been carried out examining the effects of repeated compression loading on vertebrae and how this can affect the forces required to ultimately cause compression fractures.

Brinkman et al (1988) looked into low-cycle repeated compression loading of lumbar vertebrae around the thoracolumbar junction. Low-cycle means that the number of compression cycles was limited to 5000 and the force onset frequency was 0.25Hz.

The aim of Brinkman's study was to establish how failure strength was affected (i.e. lowered) by repeated compression.

Brinkman found that at up to just 20 repeated load cycles, a number of vertebrae were found to fail at between approximately 43% and 72% of their UCS. Based on the numbers of sample vertebra which failed under various compression forces lower than their UCS, it was calculated that under repeated loading at 60 to 70% UCS, there is an approximate 9% probability of compression fracture occurring after just 10 cycles. The probability of failure will be higher if the number of loading cycles is increased to 20 and the compression force remains the same.

Due to these findings, the estimated L1 compression forces resulting from the ride motion can be considered in terms of the equivalent percentiles of full L1 UCS which they represent if it is assumed that the compression forces caused the IP's injury and if it is also assumed that the effective L1 UCS may be reduced by compression repetition.

For example if it is assumed that the compression repetitions resulted in fracture at a compression force equivalent to 60% of the IP's full L1 UCS, then the IP's full L1 UCS can be calculated. The IP's full L1 UCS can then be compared with the experimentally proven range of L1 UCS's in order to establish the following:

a) Whether the IP's estimated L1 UCS is within the range of L1 UCS typically occurring in the population;

b) Whether the L1 compression forces are sufficiently high to cause the type of injury experienced by the IP, or other types of vertebral compression fracture, when the effects of repetition are taken into account.

The IP's L1 UCS percentile represents the percentage of people that will have a similar or lower L1 UCS in the data group / population that the IP is compared with.

The IP's L1 UCS percentiles within the general population's compression strength range (as opposed to the Burst data) are calculated on the basis that the IP's injury might have occurred via a series of the bounces / ride forces causing initial minor fractures which on subsequent bounces led to the more severe burst fracture.

5.2.4 Calculations of IP's L1 UCS percentile

A series of calculations were made which determined the IP's estimated L1 UCS percentiles under various assumptions regarding effects of repetition on effective UCS. Tables 5 and 6 show these percentiles.

Table 5 shows the IP's estimated L1 UCS percentiles if the compression forces that led to injury are based on the peak forces occurring over 21 cycles or bounces and Table 6 shows the same information based on the peak forces occurring over the final 10 cycles. Two types of mean peak force are presented:

- **Peak force**

The estimated L1 compression forces are based on the mean of the absolute peak forces during 21 and 10 cycles.

- **Peak 50msec forces**

The estimated L1 compression forces are based on the mean of the average forces which occur over the highest 50msec of acceleration, i.e. for each peak, the average force over the highest 50msec of acceleration was calculated. The mean was then taken of these 21 and 10, 50msec averages.

For each mean force the tables also show a range of potential L1 UCS percentiles for the IP based on different levels of effective strength reduction due to repeated compression cycles.

Table 5. Mean estimated forces during 21 consecutive bounces & estimates of IP's L1 vertebra UCS percentile – based on a range of UCS reduction effects due to repeated compressions

Nr. bounces = 21	**Effective percentage UCS (caused by repeated compressions)**	**Full (100%) equivalent L1 UCS (N)**	**Percentile (%ile) of IP's full UCS**				
			UCS compared with burst fracture data only	**General fracture data - excluding known elderly data (>60 yrs)**		**General fracture data - including known elderly data (>60 yrs)**	
				Assuming normal distribution	**Accounting for skewness of data**	**Assuming normal distribution**	**Accounting for skewness of data**
Peak forces (N) Mean = 1296.1	F = 100%UCS	1296.1	0.9	3.0	<1.4	6.0	<1.1
	F = 72% UCS	1851.6	1.6	4.9	<1.4	9.1	6.6
	F = 43% UCS	3240.2	6.0	13.9	14.6	21.3	23.4
	F = 60% UCS	2160.2	2.2	6.3	1.8	11.3	8.6
Peak forces (N) during 50msec Mean = 1275.6	F = 100%UCS	1275.6	0.9	2.9	<1.4	5.9	<1.1
	F = 72% UCS	1822.2	1.6	4.8	<1.4	8.9	6.4
	F = 43% UCS	3188.9	5.8	13.4	14.3	20.7	22.9
	F = 60% UCS	2125.9	2.1	6.1	1.7	11	8.4

- Note: F = Compression force on IP's L1 vertebra - estimate based on ride accelerations and 50% of IP's body weight.

- Example:

  - Where F = 60% UCS in column 2, the corresponding data / percentiles are based on the assumption that the repetitive loading led to the IP's L1 vertebra failing under the compression force which was the equivalent to 60% of her normal UCS.

  - The IP's corresponding full estimated L1 UCS figure is then shown in column 3.
  - The IP's L1 UCS percentile is then calculated based on a comparison between the full estimated UCS and the respective data group.
  - The percentile figures represent the percentage of the general population who can be expected to have an L1 UCS the same as or lower than the UCS in column 3 on the corresponding row of the table

Table 6. Mean estimated forces during the final 10 consecutive bounces & estimates of IP's L1 vertebra UCS percentile – based on a range of UCS reduction effects due to repeated compressions

Nr. bounces = 10	Effective percentage UCS (caused by repeated compressions)	Full (100%) equivalent UCS (N)	Percentile (%ile) of IP's full UCS				
			UCS compared with burst fracture data only	General fracture data - excluding known elderly data (>60 yrs)		General fracture data - including known elderly data (>60 yrs)	
				Assuming normal distribution	Accounting for skewness of data	Assuming normal distribution	Accounting for skewness of data
Peak forces (N) Mean = 1391.4	F = 100%UCS	1391.4	1.0	3.3	<1.4	6.5	<1.1
	F = 72% UCS	1932.6	1.8	5.2	<1.4	9.6	7.1
	F = 43% UCS	3235.9	6.0	13.8	14.6	21.3	23.3
	F = 60% UCS	2319.1	2.6	7.3	2.3	12.5	9.7
Peak forces (N) during 50msec Mean = 1354.9	F = 100%UCS	1354.9	0.9	3.2	<1.4	6.3	<1.1
	F = 72% UCS	1881.9	1.7	5.0	<1.4	9.3	6.8
	F = 43% UCS	3151.1	5.5	13.1	14.0	20.3	22.5
	F = 60% UCS	2258.3	2.5	6.9	2.1	12.0	9.3

5.2.5 Summary of L1 UCS ranges and IP's estimated L1 UCS percentile

In the following interpretations of the IP's estimated L1 UCS ranges, the 50msec peak data in the final 10 cycle / bounce data is used (see Table 6 – data in bottom 4 rows). This data has been used in preference to the other data for the following reasons:

a) The 50msec period offers a period of force application similar to that used in a previous study, and during which fractures occurred (Lin et at, no date)

b) The effective L1 UCS reduction due to compression repetition and the related probability of failure after 10 cycles of data can be directly paralleled with the predictions made by Brinkman et al (1988) about failure after 10 compression cycles.

5.2.6 Comparison of L1 compression force with Burst fracture data

The Burst fracture data is approximately normally distributed (see Appendix 2 for details) and this indicates it is appropriate for predictions of percentiles of specific L1 UCS values based on the mean and standard deviation of the data.

The range of the IP's possible UCS's indicates the following:

- If it is assumed that the repetitive loading caused the IP's fracture to occur at 60% L1 UCS, the IP's full / usual L1 UCS would be approximately 2258N / 2.5%ile in terms of the UCS / strengths known to be required to resist burst fracture.

In the scientific studies which I have referred to for data, the lowest recorded L1 UCS for a fracture identified specifically as a burst fracture is 2542N. This means that the percentile figures associated with L1 burst fracture UCS's lower than 2542N are extrapolated values which rely on a normal distribution of the range of forces necessary to cause L1 burst fractures in the general population.

If an assumption is made of L1 UCS being reduced to 60% due to compression repetitions, then for the estimated compression forces to reach 60% of 2542N for the IP (i.e. to assume the IP had an equivalent L1 UCS / burst strength equivalent to the minimum recorded value), an increase of 7kg supported by L1 and an additional $4ms^{-2}$ (less than 0.5g) would cause a burst fracture. These figures (7kg and 0.5g) are just examples, a greater weight increase would require a lesser acceleration increase to achieve the same increase in force levels.

5.2.7 Comparison with General fracture data – excluding data from subjects >60yrs old (Hansson data)

The general fracture data is skewed positively (see Appendix 2 for details). This means that the most reliable percentile calculations for the IP's L1 UCS are based on ranked general fracture data (as opposed to percentiles which are calculated assuming a normal data distribution).

Although the IP is known to have suffered a burst fracture, this section and the next (comparisons with general fracture data) is based on the assumption that the IP's injury began as a less severe fracture caused by ride forces, which then developed into the more severe burst fracture due to subsequent bounces / compression loadings.

- If it is assumed that the repetitive loading caused the IP's fracture to occur at 60% L1 UCS, the IP's full / usual L1 UCS may have been approximately 2258N / 2.1%ile in terms of the general population.

In the scientific studies which I have referred to for general fracture data (excluding data from subjects >60yrs), the lowest recorded L1 UCS for a fracture is 2030N. The percentile figures associated with L1 UCS's lower than 2030N are all <1.4%ile because the percentiles are calculated based on rankings and the lowest known rank / data point is 2030N / 1.4%ile.

5.2.8 Comparison with General fracture data – Including data from subjects known to be >60yrs old (Hansson data)

As with the general fracture data (<=60yrs only) the data from all age groups is positively skewed (see Appendix 2). This means that IP's percentile UCS strengths are based on ranked data.

The range of the IP's possible L1 UCS's indicates the following:

- If it is assumed that the repetitive loading caused the IP's fracture to occur at 60% L1 UCS, the IP's full / usual L1 UCS may have been approximately 2258N / 9.3%ile in terms of the general population data.

In the scientific studies which I have referred to for general fracture data (including data from subjects >60yrs), the lowest recorded L1 UCS for a fracture is 1520N. This was recorded from a 72 year old female. The highest single compression force estimated to act on the IP's L1 vertebra from the bouncing motion is 1543N (i.e. 23N higher than a recorded L1 USC, albeit in an older person).

5.2.9 Significance of Weight-to-UCS ratio in determining risk of injury

It is not correct to say that simply because a person has, for example, a 1.5%ile L1 UCS, they will be at particularly high risk of injury, or at more risk for example than someone with a 10%ile UCS. Similarly, it is not correct to say that because the IP is estimated to have, for example, a 2.1%ile L1 UCS, anyone with a lower L1 UCS will also be at risk of injury. This is because the risk of fracture / fatigue is related directly to a person's body weight.

For example the person with a 1.5%ile L1 UCS and a 20%ile body weight may not be at risk of injury, whereas if that same person had a 85%ile body weight they might have an increased risk of injury due to the higher forces acting on the L1 vertebra. Similarly a person with a 10%ile L1 UCS and a 99%ile body weight may have a greater risk of injury than the 20%ile body weight person who only has a 1.5%ile UCS.

It is therefore the extent of discrepancy between body weight and L1 UCS which determines the risk of injury. The higher the relative body weight, for any specified L1 UCS, the greater the risk of injury. It is possible that this discrepancy may only be able to reach a sufficiently high level to cause injury in people with particularly low L1 UCS.

An additional complicating factor relates to how the General L1 UCS data 'population' is made up from the sum of numerous sub-groups. This arises from the assumption that generally as body weights increase the associated L1 UCS also increases (Bakker et al, 2003). For example the typical range of L1 UCS for a group of 90 to 95%ile weight males will almost certainly be higher than the typical range of L1 UCS for 10 to 15%ile females. This would therefore make it less appropriate to discuss the level of risk for a 95%ile weight male, in terms of a range of L1 UCS strength which includes data from 10%ile females.

This means that within the IP's body weight 'population' she may have had a lower percentile L1 UCS than the figure estimated based on the entire population range. The IP's body weight percentile is 70.8%ile occurring simultaneously with an estimated 2.1%ile L1 UCS. However, if we were to examine the L1 UCS range etc. in all 70.8%ile females, it is possible that the IP's L1 UCS percentile relative to that group would be lower. It is this potentially lower percentile figure that is of key relevance. The percentage of people at risk of this type of injury, on the Crazy Frogs ride, is the percentage of people with a sufficiently high percentile weight combined with a sufficiently low percentile L1 UCS.

The relationship between body weight and L1 UCS can be considered from two perspectives:

1. The overall population can be split into groups defined by a specific L1 UCS level / range. Within each of these groups there will be people with a range of body weights. The people in each group with the higher weights are those who are likely to be at greater risk of injury.

   Or alternatively,

2. The overall population can be split into groups defined by their body weight. Within each of these groups there will be people with a range of L1 UCS. The people in each group with the lower L1 UCS are those who are likely to be at greater risk of injury.

If we take the example of a weight specific 'population' e.g. people who weigh between 75 and 80kg, the range of L1 UCS will vary depending strongly on factors such as group members' age and gender (race, heredity and current / previous illness may be additional factors). Both age and gender are correlated with Bone Mineral Density (BMD) and thus bone strength, with females and older persons showing generally lower levels of BMD (Brinkman et al, 1988., Hansson et al, 1980). There may also be an added factor of Body Mass Index i.e. the ratio of weight to height. Out of two people with the same weight, one may be tall and thin and the other may be short and broad. The tall thin person will have a low BMI and the short broad person will have a high BMI. There is some evidence that people with thinner / lighter builds will be at greater risk of vertebral compression fractures due to having a narrower bone structures relative to the weight that the structures support and less muscular support around the spine. However, there is also evidence that when compression injuries do occur to more heavily built people, the injuries tend to be more severe (Edwards, 1996). These factors (BMD and BMI) are discussed later in more detail.

At this time the level of detail regarding the ratio of L1 UCS to body weight, which would be necessary to make estimate of 'within weight population' L1 UCS is not available. This means that to try and make estimates of effective safety margins, in terms of ride accelerations, it is necessary to assume that people in every weight / stature category etc. may have the same potential range of L1 UCS (i.e. the full burst / general UCS ranges). It is likely that this assumption will give relatively conservative estimates of the necessary ride acceleration safety margins, by overestimating the levels of risk to heavier individuals. An example of such a calculation of a ride acceleration safety margin during repeated bouncing motions could be based on defined parameters as follows:

Parameters: The repeated compression force occurring on the L1 vertebra of a $95^{\%ile}$ weight male should not exceed 70% of the 1.1%ile L1 IPS for the general population.

- 95%ile male body weight = 101.1kg (50% of this body weight = 50.6kg)
- Ride accelerations = 36.7ms^{-2} (mean 50msec peaks over final 10 bounces)
- Estimated force on L1 = 1857N (i.e. equivalent to a 6.6%ile L1 UCS within the general population (including older subjects).

To reduce the 95%ile males' L1 compression force to 1520N / 1.1%ile of the general population (and the lowest recorded compression force to have caused L1 to fracture), the peak z and y-axis vector accelerations would need to be reduced to 30.04ms^{-2} i.e. 3g accelerations. To include an additional safety margin which assumes a 70% effective L1 UCS due to repetitions, this figure would need to be lowered to 21.03ms^{-2} i.e. 2.2g peak vector accelerations.

Load cases supplied by the ride manufacturer indicate that the peak z-axis accelerations on the ride may be as high as 4.65g in the z-axis (load case 1) and I measured peak accelerations as high as 4.11g in the z-axis. If this lower acceleration threshold of 2.2g were to be adopted, it may be necessary to install a limiting mechanism on the pressures used to drive the ride arms. If further information became available on 'within population' L1 UCS ranges, this 2.2g figure could then be revised in line with that new information.

6.0 IP’S L1 UCS PERCENTILE ESTIMATES BASED SOLELY ON Z-AXIS ACCELERATIONS

The estimations of the IP’s L1 UCS percentiles in the previous sections are based on the assumption that passengers will lean into the ride slightly to counteract / eliminate a lateral shear force on their neck / upper body, and will thus experience the combined vector of the z and y-axis g-forces axially though their spine.

In order to determine the potential significance of that assumption, IP’s L1 UCS percentiles were calculated in exactly the same way as described previously, but based solely on the z-axis accelerations. Appendix 3 contains tables which give estimates of the IP’s L1 UCS based on the assumption that only the z-axis accelerations contributed to the vertebral compression force.

Comparisons between the estimated L1 compression forces indicate that the peak forces over 21 bounces and 10 bounces are only 4% lower if they are based solely on z-axis data compared to the combined z and y-axis vector acceleration data.

The 4% lower forces result in a decrease in the IP’s estimated L1 UCS percentile of between 0.2% (based on the burst fracture data group) and 0.3 to 0.5% (based on the ranked percentiles of the General data group <=60yrs and the General data group >60yrs incl. respectively). Such differences are not large enough to significantly affect the conclusions about risk levels which have been based on the z and y-axis vector force.

7.0 GENERAL DISCUSSION : FACTORS EFFECTING OVERALL RISK OF INJURY

7.1 Lack of awareness of personal proneness to vertebral compression injury

A number of factors determine the UCS for a vertebra. These factors are primarily; a) bone mineral density and b) the load bearing cross-sectional area of the vertebral endplate (Brinkman et al, 1989). A key issue is that these are both factors which people generally have no knowledge or awareness of, regarding themselves or the people they accompany on rides etc. This means that people whose UCS is within the lower percentiles of the general populations UCS range may use the ride and thus be exposed to a risk of injury which is dependent on their upper body weight.

7.2 Bone Mineral Density

Low Bone Mineral Density (BMD) is associated with lower UCS (Hansson, 1980., Brinkman, 1988). Typically, older people will have reduced levels of BMD and this is one of the key reasons why a greater percentage of the population of an L1 UCS data group which included subjects aged >60 may have a risk of experiencing a vertebral fracture during bouncing motions on the Crazy Frogs ride. Females have also been shown to have a lower mean BMD than males (Hansson et al, 1980). These correlations mean that these groups, the elderly and females, are generally likely to be at more risk of injury than for example younger people and males.

Although lower UCS is closely associated with lower BMD, there is no indication from the research studies that any of the subjects were known to have bone mineral deficiencies i.e. at the time of their death none of the subjects were aware / had been diagnosed as having any clinical conditions directly associated with low BMD (such as osteoporosis). This means it is possible that people of any age with low BMD (and thus low vertebra compression strengths) could use the ride without being aware of their relatively higher risk of injury from axial vertebral compression forces.

7.3 Body Mass Index

Amongst the general population there will be some individuals who have an uneven weight distribution where more than 50% of the body weight is supported by the L1 vertebra (and thus adjacent / consecutive thoracolumbar vertebra also support a higher percentage of body weight). Because this will increase the relative compression force acting on the L1 vertebra, without the strength of their vertebrae being similarly increased, people who possess this type of top-heavy body shape may be at greater risk of compression injury resulting from the forces during bouncing motion.

On the other hand there is evidence that tall and thin people with lower BMI have been associated with increased levels of compression fractures in studies of air-crew ejection injuries (Edwards, 1996). This has been suggested as being due to their having relatively less muscular support of the spine compared with more heavily built people.

It is also possible that these taller thin people will tend to have a narrower bone structure which for a given compression force will result in greater pressure on the vertebral endplate and thus a greater risk of fracture. If a person puts on weight thus increasing their BMI, providing that the weight is evenly distributed then the skeletal structure may adapt, to some extent, to the weight (increased body fat and skeletal muscle development has been associated with increases in bone density / strength (Stewart et al, 2002., Tarquini et al, 1997)). However, there may also be limitations placed on this effect due to the importance of the cross-sectional area of the vertebral endplate (in fully developed adults the vertebral cross-sectional area will not change as body weight changes) (Brinckmann et al, 1989). This would support the case that if more than 50% of the additional weight is supported by the L1 vertebra (i.e. a top-heavy build), any corresponding increase in bone strength may not be sufficient to offer similar relative strength / support as prior to the weight gain. This also means that in people with extremely high BMI the risk of vertebral compression injury may be increased where the body weight related increases in bone strength can no longer compensate fully for the increase in weight. However it is not possible to say based on the information available, what the 'safe' upper or lower BMI thresholds may be. It is simply necessary to note that people at the extremes of BMI are more likely to have a greater discrepancy between their L1 UCS and their weight.

It should also be noted that females tend to have narrower vertebra than males (and thus vertebra with smaller surface areas) and that post-menopausal women show the most marked effects of the higher body fat : higher BMD relationship.

7.4 Additional Risk Factors

7.4.1 Long term shock / vibration exposure & Spine diseases

People who have been previously exposed to whole-body vibration in a work environment or shocks during vehicle use may be at greater risk of vertebral compression injury due to the accelerations / g-forces on the Crazy Frogs ride. There is a belief that prolonged exposure with healing periods in between can lead to thickenings in the soft tissues surrounding the vertebrae which may reduce the nutrient flow to the intravertebral discs, and thus reduced tolerance to high-acceleration events (Brinkman et al, 1988).

People with tumours in their vertebrae and people with localised bone infection which reduces BMD / bone density, tolerance to strain etc. are all likely to have significantly more risk of injury than people with healthy spines.

7.4.2 Inappropriate cushioning

Little cushioning was present on the Crazy Frogs' seats, however the research literature suggests that providing cushioning is not a straightforward way of reducing the risk of vertebral injury. A cushion is not likely to reduce the peak forces that act on the spine. There is also a risk that if the cushion 'bottoms-out' the initial acceleration of the seated person up to that point in time will not have been as rapid as the seat itself. This could result in the person having to 'catch-up' with the seats' acceleration levels leading to a more powerful and more rapid jerk or rate of change of g than would occur on an uncushioned seat or a properly cushioned seat (Hodgson et al, 1963). These risks associated with cushioning are identified because they may be relevant to any future alterations to the ride, or other rides with similar motion patterns.

7.4.3 Effects of Posture

Posture is believed to have a significant bearing on the effects of g-force accelerations in the vertical plane. It has been demonstrated that as the spine is flexed forwards the turning moments sustained by the vertebrae are increased, and this is likely to result in higher compression forces between the vertebrae (King et al, 1970). Throughout the literature on spine injury / fracture there are consistent references to this detrimental effect which spine flexion can have on the vertebrae's tolerance of compression forces.

When the ride is being rotated in reverse passengers may be more prone to slumping forwards (spine flexion) over the handrail, either consciously or inadvertently as a result of relaxing trunk / back muscles. Similar reduction in posture control may result from rapid / repeated onset of accelerations, this effect is explained in more detail in the next section.

7.4.4 Jerk / rate of change of g

The rate of change of g / acceleration, known as jerk (ms^{-3}), is a factor that has been linked to the rate of vertebral compression / strain (Hodgson et al, 1963). The jerk levels that have been previously studied in relation to injury are significantly higher than those which were measured on the Crazy Frogs ride and involved much higher g levels. Jerk is typically considered in terms of its effects on people's ability to anticipate movements, e.g. side to side movements on a Rollercoaster. If acceleration builds up rapidly it is harder to adjust the body's posture to counter the forces which could result in injuries such as whiplash.

Jerk values were calculated over periods of between 0.025 and 0.375 seconds (25 to 375 milliseconds). Figure 5 shows the mean and maximum jerk levels for each duration of calculation. The graph shows clearly that over shorter periods the jerk levels are higher. This is due to the combination of a relatively low sample rate and the short durations of jerk steps.

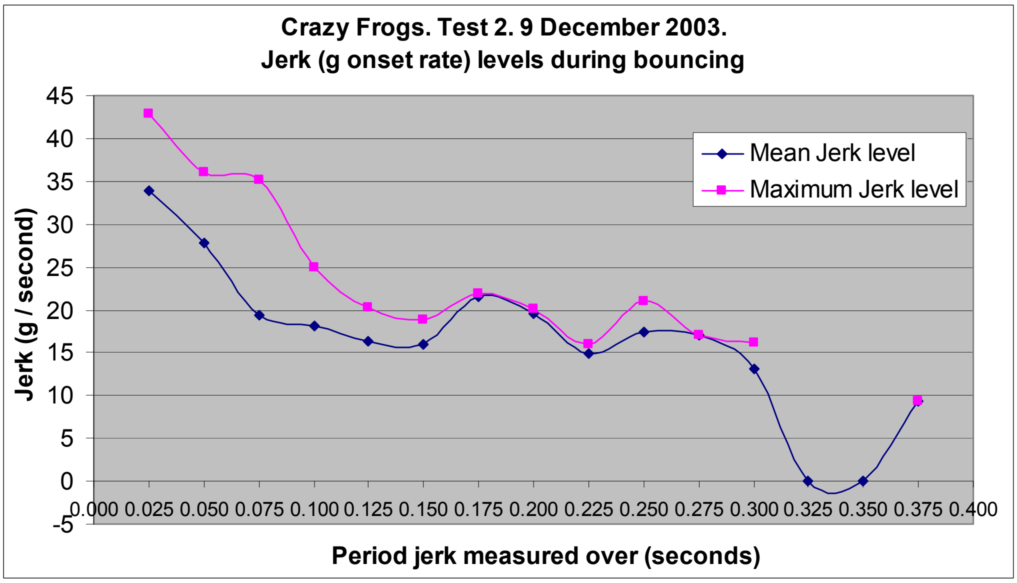

Figure 5. Range of jerk levels depending on measurement period.

The jerk levels during bounces ranged up to 42.8 g per second.
The average maximum jerk across the 22 bounce peaks was 23.05 g per second (Std. Dev = 9.42).

In the Crazy Frogs ride the onset of peak acceleration over short periods during bouncing motions (over approximately 200ms from 0g to peak g) may mean that some passengers' muscular systems (back and stomach muscles) may not respond in time to provide effective support capability over successive bounces. The average reaction time (signal detection and action decision period) to a stimulus is approximately 200ms (0.2 seconds), or possibly 160ms for a primarily touch stimulus. The minimum additional time to complete a muscular response is approximately 300ms (McCormick, 1976). This results in a total stimulus-response time of approximately 460 to 500ms which is longer than the period that a person would have to respond before a peak g-force (i.e. a maximum compression force) occurred during bouncing motions.

Alternatively the muscles may begin to fatigue after a number of bounces, again possibly resulting in a loss of support. Given that muscular support is also necessary to maintain an upright posture during upwards accelerations / downwards decelerations, any reduction in that muscular support may cause inadvertent spine flexion (slumping forwards) which will increase the risk of vertebral compression injury.

To determine the effects of jerk on the compression of vertebrae it would be necessary to determine the energy which is transmitted through a vertebra as a result of a jerk. This could be calculated as total energy or as power (energy / joules per second.). In this respect an increase in acceleration can be considered in terms of how powerful it is, with faster rates of change in acceleration being the more powerful jerks. Because jerk determines the rate of increase of compression force, jerk levels may affect the rate of strain (i.e. millimetres of deformation per second) which is experienced by vertebrae under compression. This may, in theory, make it possible to compare the estimated energy imparted on the L1 vertebra with the strain energy limits of L1 vertebrae. In this investigation, this comparison was not possible for two reasons;

- Cushioning effects of soft tissue between the spine and the seat are likely to affect the rates of jerk / compression force onset actually experienced by the L1 vertebra. For example the cushioning may delay the initial force increase thus causing a more powerful but shorter duration jerk leading up to the peak force level during the middle / towards the end of the period of acceleration.

- Insufficient scientific data could be found to compare a range of L1 maximum energy (to fracture) with possible energy predictions based on the Crazy Frog g-forces.

The negative-g troughs in the g-force data (and the jerk levels) indicate that some separation may occur between the passenger and the seat during the bouncing motions. There are very brief increases in measured g-force immediately following these negative accelerations. These are likely to be the points at which the dummy came back into contact with the seat after a period of acceleration. These did not coincide with the main acceleration peaks and will therefore not have contributed to the recorded peak g-accelerations. However, the dummy which the accelerometer was attached to was a passive object, it is possible that a person may slightly delay their return to the seat via muscular effort in their arms. If this happened, the person's return to their seat may coincide with the peak upwards acceleration of the seat, thus increasing the spinal compression forces which they experience. It is therefore advisable that people should not be able to lift off their seats as a result of their upwards acceleration.

8.0 COMPARISON OF CRAZY FROGS' G-FORCES WITH OTHER RIDES' G-FORCE DATA

The risk level associated with g-forces on different rides depends on the design of the ride itself. A higher z-axis g-force may be more acceptable if it builds up slowly enough for passengers to adjust their posture or if their upper body weight is well supported by their seating. However, if the same force is onset rapidly, there is limited upper body support from the seating and the force is repeated, then there is likely to be an increased risk of injury. This means that the peak forces on the Crazy Frogs ride may be more critical those found on some other rides, because the force onset is relatively fast, passengers are not necessarily supported by the seat back, and the force is repeated up to 20 times.

HSL has a database of g-acceleration gathered from 24 ride assessments / investigations (Described in HSL report ERG/01/10). The database shows that out of 24 rides where z-axis g-forces are known, only 5 showed maximum peak z-axis g-forces greater than the peak 4.11g measured on the Crazy Frogs ride. The peak z-axis g-force measured on three separate log-flumes (i.e. generated by the single drop) was approximately 3g. In all of the log-flumes this was a single peak force event. The peak z-axis g-forces measured on 3 roller coasters were generally found to be between 2.5g and 3.5g with a single maximum peak of 6.4g on one of the coasters.

The Tagada is one of the only other rides in operation which generates a 'bouncing' motion with some similarities to the Crazy Frogs ride. The Tagadas peak z-axis g-force measured by HSL was 2.5g (this was the highest peak z-axis g-force measured from 3 different Tagadas). In addition to the peak z-axis g-force being significantly lower than the Crazy Frogs ride, the spinning motion of the Tagadas disc during bouncing means that passengers will not experience as many consecutive peak compression forces.

This comparison of the g-forces on the Crazy Frogs ride with other ride g-forces shows that as well as the conditions of motion (repetition, fast force onset, seat back potentially unused) making the effects of any z-axis g-forces more critical than many other rides, the forces themselves are also higher than 80% of other rides which have had g-force data recorded by HSL. In all but one of the cases where other rides z-axis g-forces are higher than 4.11g, the peak forces are not part of repetitive bouncing patterns as found on the Crazy Frogs ride.

The only ride that has been investigated by HSL which showed a similar potential level of risk was a Mont Blanc type ride which was assessed by HSL in 1998 following concerns over passenger ejection. The peak z-axis g-force was found to be 5.6g and it was advised in the report (HSL report EWP/98/03) that the force may cause potentially dangerous shock loading of the spine. Although the peak z-axis g-force was found to be 5.6g, the majority of the peaks were approximately 4g or less with a number of peaks under 3g. The report (EWP/98/03) advised that the peak z-axis g-forces should be reduced by discontinuing the use of controls which increased the z-axis g-forces, and that the operator should look into revising the ride control system.

9.0 CONCLUSIONS

Certain types of ride motion may expose passengers to greater risk of injury. The g-force data indicates that rapid bouncing motions (measured at approximately 1.2 Hz) produce the highest g-forces and the highest rate of change of g.

9.1 Conclusions of comparisons between IP's L1 UCS and known L1 UCS

The following conclusions are based on estimates of compressive forces on the IP's L1 vertebra made under the following assumptions:

- L1 vertebra supports 50% of IP's weight (36.95kg at time of injury).
- Z and y-axis vector accelerations (primarily downwards deceleration and upwards acceleration) cause the axial compression

The conclusions are also based on the forces estimated to have been generated during the 10 final peaks of the bouncing pattern. Although there is no significant effect on the overall conclusions if the average peak forces over 21 of the bounces are used instead.

- **IP's estimated L1 UCS if repetitive loading effects on strength not considered**

Based on the estimated compression levels caused by the z and y-axis vector acceleration during a single event, for the IP's L1 vertebra to suffer a burst fracture then the IP's L1 UCS would be in the weakest 1% of the general adult population (1354N or less).

- **Significant effects of repeated compressive loading of vertebrae**

Given the scientific evidence available it is appropriate to consider that the frequent repetition of the compression forces will reduce the effective UCS of vertebrae such that after 10 repeated cycles, a compression force equivalent to just 60% to 70% UCS is sufficient to result in a 9% probability of the vertebrae fracturing.

- **IP's L1 UCS based on comparison with burst fracture data**

When the effective UCS is reduced to 60% by repetitive loading, and 60%UCS is presumed to be the force that led to the IP's injury, the IP's equivalent full UCS would be approximately 2258N / 2.5%ile, i.e. the IP's L1 UCS would be in the weakest 2.5% of the known strengths relating to burst fractures.

- **IP's L1 UCS based on comparison with general L1 UCS data**

If the IP's full estimated UCS (again assuming a 60% effective reduction) is compared to the UCS from a general population (where various types of fractures occurred) the IP's L1 UCS would be within the lowest 2.1% to 9.3% of the population; with the higher percentile calculation being based on a comparison group which includes data from adults aged >60yrs. This conclusion assumes that over consecutive compressions, minor fractures would develop into more severe burst fractures.

- **Importance of weight / L1 UCS discrepancy**

It is clear that the level of risk caused by z-axis accelerations for an individual with a specific strength of L1 UCS is determined by the amount of discrepancy between the person's L1 UCS and their weight (i.e. the weight supported by the L1 vertebra). It is likely that in general people who are heavier will tend to have higher L1 UCS (and people can thus be considered as belonging to a weight 'population' with its own range of L1 UCS). This would largely explain why there have not been significantly more catastrophic fractures occurring to passengers on the ride. However, at this time there is not sufficiently detailed data available to quantify the level of risk across the general population based on the likelihood / occurrence of sufficiently high weight / L1 UCS discrepancies i.e. it is not possible to say that a particular percentage of people will be at risk of vertebral compression injury due to the bouncing motions.

- **Example calculation of an adequate safety margin**

In order to develop an adequate margin of safety, it is therefore necessary to assume that heavier people / people with top heavy builds may have L1 UCS within the lower 1% of the general populations range of L1 UCS. An example calculation of a safety margin which assumes that there will be no passengers with a discrepancy as high as 95%ile male weight / 1.1%ile UCS indicates that if compression repetitions are assumed to reduce effective L1 UCS by 70%, the peak bounce accelerations should not exceed 2.2g. Based on the same example person, where repetitive compression is not significant then peak forces of 3g would be deemed to create an adequate safety margin.

- **Lowest recorded L1 UCS**

The highest single compression force estimated to act on the IP's L1 vertebra from the bouncing motion is 1543N which is 23N higher than the lowest recorded L1 USC in the scientific data which is referred to in this report. It is noted that the 1520N L1 UCS was recorded in an elderly female.

- **Lowest recorded force causing burst fracture**

The lowest recorded force to cause an L1 Burst Fracture is 2542 (based on an increment of 538N added to a 2004N T12 vertebras Burst fracture strength). If a 60% reduction in effective UCS is assumed, then if the IP had an L1 UCS of 2542N, an additional 7kg weight supported by L1 and an additional $4ms^{-2}$ would lead to an estimated 9% probability of a burst fracture occurring.

- **Compression forces based solely on z-axis data**

If the estimates of L1 compression forces are based solely on z-axis accelerations, there is a slight decrease in estimates of IP's L1 UCS however the decrease does not alter the conclusions.

9.2 Additional conclusions

- **G-forces in comparison with other rides**

The peak z-axis accelerations of a seated passenger during bouncing are higher than a number of rides from which HSL has recorded g-forces. However, certain rides have exhibited similar or higher peak accelerations. It is likely that the repetitive accelerations / compressions create a unique level of injury risk associated with the peak g-forces.

- **Bone Mineral Content**

There is no indication from the scientific literature that for someone to have a high-risk discrepancy between their weight and their L1 UCS their BMD needs to be sufficiently low that they are likely to have been diagnosed as having a clinically low BMD (linked to a condition such as osteoporosis, spinal tumour, drug use etc.).

- **Build / Physique**

People at the extremes of Body Mass Index (BMI) may be at greater risk of vertebral compression fracture.

  - **Tall and Thin people (Low BMI)**

    There is evidence that tall and thin people may be more at risk of vertebral fracture injury than shorter and broader people. It is inferred that tall / thin people may therefore have a greater discrepancy between their weight and their vertebral UCS'. This may be due to thinner people having a narrower bone structure / smaller vertebral endplate cross-section combined with a substantial weight still being supported. It may also be due to their having relatively less muscular support around the spine. There is also some association between body fat levels (i.e. generally lower BMI) and bone strength.

  - **Heavy / Top heavy people (potentially high BMI)**

    Although it is possible that heavier people will generally have higher L1 UCS levels, the effects which may create this balance may only be effective up to a certain BMI, beyond which the fixed vertebral cross sectional area limits further strength increases. Based on this, people who develop a particularly high BMI may be at increased risk of injury. Similarly if potential vertebra strengthening effects are based on body weight / body fat increase, the strengthening mechanisms may not be able to detect when the weight distribution is uneven (e.g. 'top-heavy'). This may create a widening discrepancy between a persons weight and their L1 UCS, thus increasing the risk of injury. Finally, there are indications that when a heavier person does sustain a vertebral fracture the injury tends to be more severe.

- **Posture / Trunk Flexion / Effects of Jerk**

Trunk flexion (leaning forwards) is a posture that can be adopted during bouncing and it is likely to increase the risk of vertebral compression fracture. The high frequency of the bounces, the rapid onset of peak forces and the cumulative number of bounces could fatigue the trunk / back muscles necessary for maintaining an upright posture. Alternatively these muscles may not have time to react to force onset. Either of these scenarios could result in inadvertent trunk flexion and thus an increased risk of injury.

- **L1 UCS in children**

The calculations of injury risk in this report are based on vertebral compression strengths of adults. Although one would expect L1 UCS to be fairly high relative to body weight in children (due to their having relatively high BMD because of their young age) I have not found any definite evidence to that effect. It is also important to note that there are medical conditions which can occur at young ages such as Anorexia Nervosa or Leukaemia which may reduce Body Mass Index to an extremely low level, which may also reduce the BMD / compressive strength of vertebrae relative to weight thus potentially increasing the risk of vertebral compression injury.

- **Summary of groups at greatest risk of vertebral compression injury**

There are certain groups of the population who will be particularly at risk of vertebral compression injury due to the bouncing motion of the Crazy Frogs ride. These are as follows:

  - Older people
  - Females
  - Taller people (heavy but with narrow bone structure / low BMI)
  - People with uneven / top-heavy weight distributions
  - People with bone mineral deficiency conditions such as osteoporosis
  - People suffering from other spinal / bone disease (tumours etc.)
  - People with conditions reducing body weight / nutrient metabolism.

10.0 RECOMMENDATIONS

10.1 Ride accelerations / Motion properties

- The high-frequency bouncing motion should be restricted so as to avoid generating peak z-axis accelerations at the levels that were recorded (up to 4.1g in z-axis) because they present a significant risk of vertebral injury to a proportion of potential passengers.

- If the bouncing motion continues to be used, the peak z-axis accelerations should be reduced to a level which provides an appropriate margin of safety. The reduced accelerations should take into account the potential for heavy people to have low L1 UCS, and the reduction in effective L1 UCS caused by repetitive compressions.

- To further reduce the risk of vertebral compression injuries, the bouncing motion should not be used for sustained periods. For example no more than 10 consecutive bounces should occur.

- The ride should not be run in reverse for long periods (and the reverse motion should not be combined with the bouncing motion). Running the ride in reverse may cause people to consciously or inadvertently lean forwards thus placing their spine in flexion.

- All of the ride's motions / motion patterns should be considered when determining an appropriate safety margin for accelerations. This report focuses on the higher acceleration bounce motion, however, relatively high repetitive accelerations also occur during the slower wave motions. Assessments of accelerations / forces should concentrate on the outer seats because these will show higher accelerations than the inner seats.

10.2 Signs, Instructions and Filtering Passengers

Reduction in bounce accelerations is considered necessary because the key physiological factors which put individuals at increased risk would often be very difficult for people themselves, let alone an operator, to identify and use to select out 'at risk' individuals. However, the following steps could be taken to enhance a safety margin created by a reduction in bounce accelerations:

- People with <u>any</u> history of back injury / back problems should be warned not to use the ride. Particular attention should be paid to wording and conspicuity of signage – perhaps even introducing a warning on the cars themselves as a backup to other signs.

- Signs should also warn tall people, female adults (aged 40+) and any older people (e.g. over 60 / 65s) of the nature of the ride and their increased risk of injury.

- Operators should remain aware of the people who are getting on the ride; for example they could warn any particularly elderly people or people who appear to be stooped / hunched, as if through spinal musculoskeletal problems, that the ride is vigorous and may be unsuitable for them.

- Passengers should be advised not to lean forwards and to try to remain upright. This may require signs such as ‘Do not lean forwards over lap bar and handrail’.

10.3 Restraint modifications

- Although restraint modifications should be considered only in addition to a reduction in peak accelerations, preventing passengers from moving into higher risk postures (spine / trunk flexion) is likely to reduce the risk of injury. Interlocked inertia-reel chest strap restraint system or over shoulder restraints could offer this facility. However, if the ride accelerations are sufficiently reduced, there should be little need for restraint modifications.

- Extra padding could be fitted to the lap-bar to help secure legs against the seat and prevent passengers lifting off the seat and thus avoid the risk of the impact of passengers returning to their seats coinciding with the peak upwards accelerations of the seat itself.

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APPENDIX 1. L1 COMPRESSIVE STRENGTH DATA

L1 Compressive Strength (N)	Study	Defined value measured
7146.7	Kazarian et al (1977)	Compressive breaking loads
10662.2		
3916		
11567	Hutton et al (1982)	Compressive strength
8553		
5215	Hutton et al (1979)	Breaking load (load at which force / deflection shows first peak - vertebral body is damaged beyond repair
4604		
7010		
2933		
2030		
9480		
3390	Shirado et al (1992)	Load at burst fracture (failure load)
3303	Shono et al (1994)	Failure loads (forces recorded as causing burst fractures)
12535		
10550		
9950		
8550		
8650		
8350		
8300		
8250		
7850		
7650		
7650		
7550		
7450		
6650		
6650		
6550		
6350		
5550		
5550		
5550		
5050		
4650		
3650		
2542	Lin (no date)	Failure load (plus increment)
7063.2	Ruff (1950)	Breaking load (load at which force / deflection shows first peak - vertebral body is damaged beyond repair
8240.4		
7848		
7848		
8829		
9200	Brinkman (1988)	Ultimate Compressive Strength
4200		
3500		

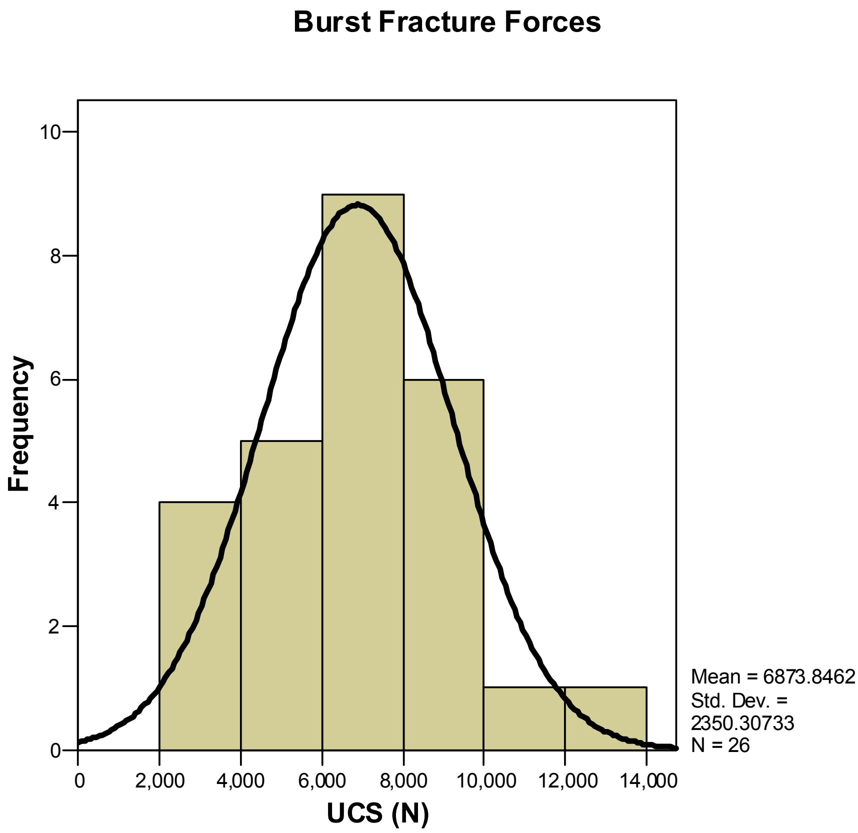
L1 Compressive Strength (N)	Study	Defined value measured
4500	Brinkman (1988)	Ultimate Compressive Strength
4500		
7500		
6100		
3900		
6000		
2800		
7900		
3100		
5300		
2900		
3384	Hansson et al (1980)	Ultimate compressive load / compressive strength. The load measured at fracture
2550		
5640		
6278		
4022		
5199		
4120		
3727		
6180		
3531		
2550		
2452		
3531		
2746		
4218		
4512		
3825		
2452*		
4414*		
2256*		
1716*		
3237*		
3041*		
1863*		
3335*		
3531*		
3335*		
1618*		
1716*		
3727*		
2060*		
3727*		
1863*		
1520*		

* = Data from adults known to be older than 60yrs

Lin (no date) data (2542N to create burst fracture) is based on an increment of 538N added on to a 2004N failure load for the T12 vertebra. This increment of 538N is based on the average intervertebral strength increases in the data collected by Ruff (1950). Typically each vertebra has a greater compressive strength than the one above it. The 538N increment is conservative (i.e. it may provide slightly higher than actual L1 burst fracture strength) compared with the lower 380N increments proposed by Brinkman (1988).

APPENDIX 2. DISTRIBUTIONS OF L1 UCS DATA GROUPS

Forces specifically leading to L1 Burst fracture

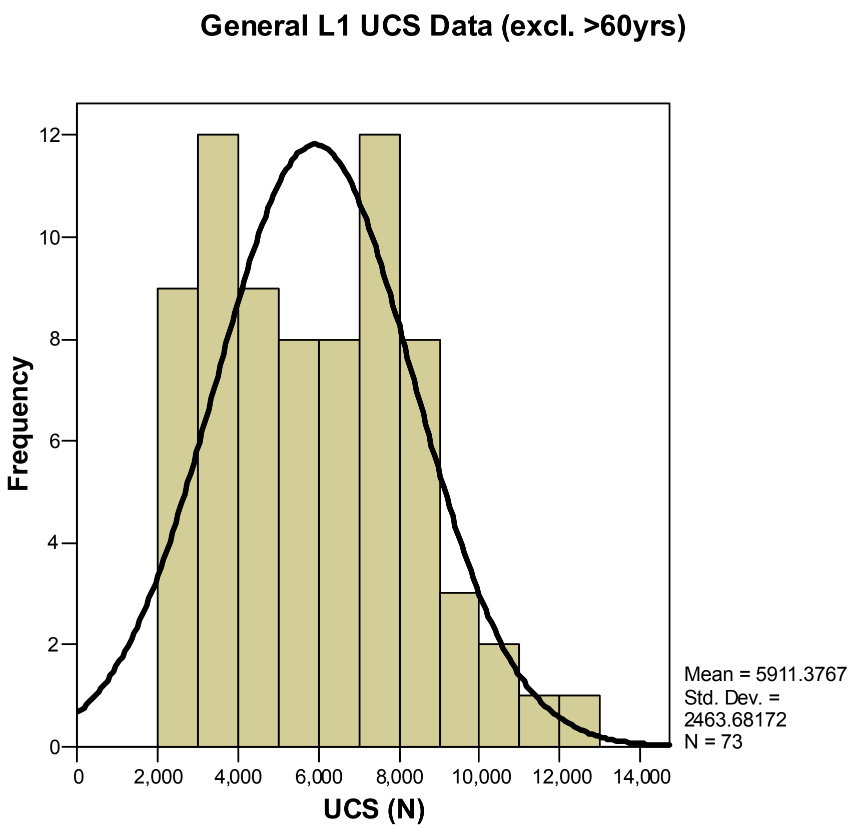


Statistics

UCS

N	Valid	26
	Missing	0
Mean		6873.8462
Median		7050.0000
Mode		5550.00
Std. Deviation		2350.30733
Skewness		.183
Std. Error of Skewness		.456
Kurtosis		.182
Std. Error of Kurtosis		.887
Percentiles	25	5425.0000
	50	7050.0000
	75	8312.5000

General L1 Compressive Strength Data (excluding subjects known to be >60yrs)

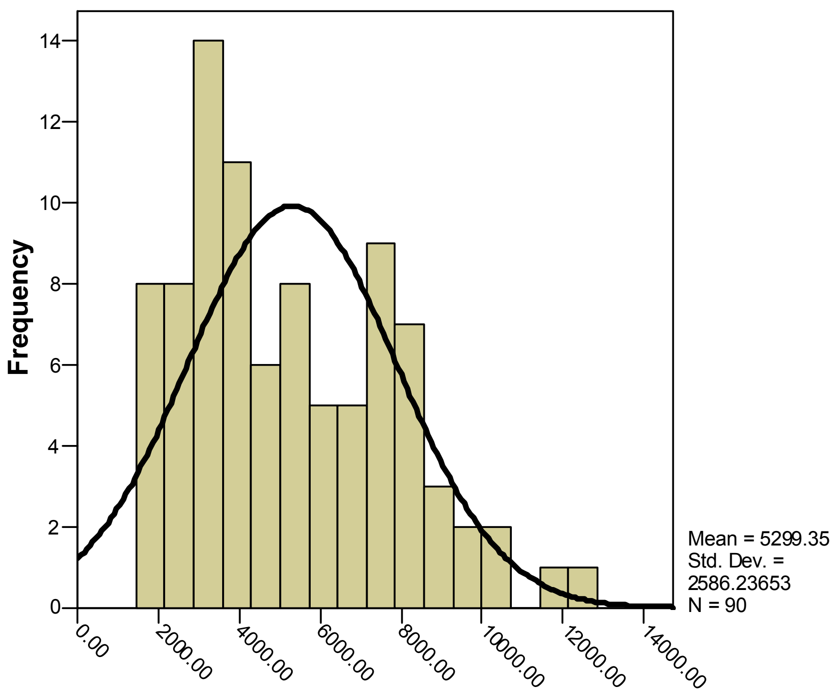


Statistics

UCS

N	Valid	73
	Missing	0
Mean		5911.3767
Median		5550.0000
Mode		5550.00
Std. Deviation		2463.6817 2
Skewness		.471
Std. Error of Skewness		.281
Kurtosis		-.476
Std. Error of Kurtosis		.555
Percentiles	25	3776.0000
	50	5550.0000
	75	7848.0000

General L1 Compressive Strength Data (including subjects known to be >60yrs)



Statistics

UCS

N	Valid	90
	Missing	0
Mean		5299.3500
Median		4558.0000
Mode		3531.00(a)
Std. Deviation		2586.23653
Skewness		.598
Std. Error of Skewness		.254
Kurtosis		-.409
Std. Error of Kurtosis		.503
Percentiles	25	3327.0000
	50	4558.0000
	75	7512.5000

a Multiple modes exist. The smallest value is shown

APPENDIX 3. IP’S L1 UCS PERCENTILES ESTIMATED <u>SOLELY BASED ON Z-AXIS G-FORCES</u>

Table 7. Mean estimated forces during 21 consecutive bounces & estimates of IP’s L1 vertebra UCS percentile – based on a range of UCS reduction effects due to repeated compressions

Nr. bounces = 21	**Effective percentage UCS (caused by repeated compressions)**	**Full (100%) equivalent L1 UCS (N)**	**Percentile (%ile) of IP’s full UCS**				
			UCS compared with burst fracture data only	**General fracture data - excluding known elderly data (>60 yrs)**		**General fracture data - excluding known elderly data (>60 yrs)**	
				Assuming normal distribution	**Accounting for skewness of data**	**Assuming normal distribution**	**Accounting for skewness of data**
Peak forces (N) Mean = 1244.1	F = 100%UCS	1244.0	0.8	2.9	<1.4	5.7	<1.1
	F = 72% UCS	1727.7	1.5	4.4	<1.4	8.4	5.8
	F = 43% UCS	2893.0	4.5	10.8	12.3	17.6	20.5
	F = 60% UCS	2073.3	2.0	5.9	1.5	10.6	8.1
Peak forces (N) during 50msec Mean = 1219	F = 100%UCS	1219.6	0.8	2.8	<1.4	5.6	<1.1
	F = 72% UCS	1693.9	1.4	4.3	<1.4	8.2	5.5
	F = 43% UCS	2836.4	4.2	10.6	11.9	16.9	20.1
	F = 60% UCS	2033.8	1.9	5.7	1.4	10.3	7.8

Table 8. Mean estimated forces during the final 10 consecutive bounces & estimates of IP's L1 vertebra UCS percentile – based on a range of UCS reduction effects due to repeated compressions (again – based solely on the z-axis g-forces, not the z and y-axis vector forces)

Nr. bounces = 10	**Effective percentage UCS (caused by repeated compressions)**	**Full (100%) equivalent UCS (N)**	**Percentile (%ile) of IP's full UCS**				
			UCS compared with burst fracture data only	**General fracture data - excluding known elderly data (>60 yrs)**		**General fracture data - excluding known elderly data (>60 yrs)**	
				Assuming normal distribution	**Accounting for skewness of data**	**Assuming normal distribution**	**Accounting for skewness of data**
Peak forces (N) Mean = 1336.5	F = 100%UCS	1336.5	0.9	3.1	<1.4	6.2	<1.1
	F = 72% UCS	1856.2	1.6	4.9	<1.4	9.1	6.6
	F = 43% UCS	3108.1	5.4	12.8	13.8	19.8	22.3
	F = 60% UCS	2227.4	2.4	6.7	2.0	11.8	9.1
Peak forces (N) during 50msec Mean = 1300.9	F = 100%UCS	1300.9	0.9	3.0	<1.4	6.0	<1.1
	F = 72% UCS	1806.9	1.6	4.7	<1.4	8.8	6.3
	F = 43% UCS	3025.4	5.0	12.1	13.2	19.0	21.6
	F = 60% UCS	2168.2	2.2	6.4	1.8	11.3	8.7