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ORIGINAL ARTICLE

Investigation of the force associated with the formation of lacerations and skull fractures

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Abstract Post-mortem examination is often relied upon in order to determine whether a suspicious death was natural, accidental, suicidal or homicidal. However, in many cases the mechanism by which a single injury has been inflicted cannot be determined with certainty based on pathological examination alone. Furthermore the current method of assessing applied force relating to injury is restricted to an arbitrary and subjective scale (mild, moderate, considerable, or severe). This study investigates the pathophysiological nature of head injuries caused by blunt force trauma, specifically in relation to the incidence and formation of a laceration. An experimental model was devised to assess the force required to cause damage to the scalp and underlying skull of porcine specimens following a single fronto-parietal impact. This was achieved using a drop tower equipped with adapted instrumentation for data acquisition. The applied force and implement used could be correlated with resultant injuries and as such aid pathological investigation in the differentiation between falls and blows. Experimentation revealed prevalent patterns of injury specific to the reconstructed mechanism involved. It was found that the minimum force for the occurrence of a laceration was 4,000 N.

Keywords Blunt force trauma · Head injuries · Laceration · Applied force

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Introduction

Damage to skin

A wound has been defined as any 'damage to any part of the body due to the application of mechanical force' [1]. Excessive mechanical forces on bodily tissues can, in some cases, cause lacerations to the affected area and the pattern of laceration to be determined by the structural and biomechanical properties of the skin [2]. The organised structure of skin serves as the primary protective barrier from the external environment, including mechanical trauma such as friction, impact, pressure, cutting and shearing [3, 4]. This protection is achieved though the skin's non-homogenous, non-linear, anisotropic viscoelastic properties [5]. Skin can be divided into two main structured layers; the outer epidermis and the underlying dermis. Skin thickness normally ranges from 4 to 0.5 mm, and thus has varying mechanical properties with varying location [6]. The mechanical resistance of skin is mainly concentrated in the dermis in the form of a matrix of collagen and elastin fibres [5, 7]. Collagen and elastin fibre arrangement in the papillary dermis is fine and vertically orientated while the reticular dermis beneath shows a thicker, coarser network with a longitudinal arrangement. This causes creased tension lines known as 'Langer lines', first described and plotted by Karl Langer [8], and are a key intrinsic factor required for consideration during the forensic interpretation of lacerations [9, 10].

Skin is in a state of unequal biaxial tension over the body, varying with respect to movement of joints and volume of mass under the skin [11–14]. When skin is relaxed, the collage and elastin fibres are unordered. When a load is applied to skin, it responds by dissipating the energy via its viscous component. The skin's load response is further explained by Young's modulus, or the stress–strain curve [3, 11]. Applied strain initially causes the elastin fibres to carry the load and collagen fibres remain unordered. As the strain increases, collagen gradually aligns in the direction of the load, where at the highest level of strain the collagen carries almost all the load until it finally fails. Collagen fibres aligned in the direction of an applied force have been shown to fail at a strain of 5–6% and strengths in the range of 147–343 MPa, varying with location on the body [6, 12, 15, 16].

The testing of skin can estimate values such as ultimate tensile strength, elasticity and density, while animal models provide information of the skin's response to external forces of stretch, shear, torsion, compression and indentation [3, 11, 15]. The kinetic energy absorbed by the surface on impact causes the skin to deform. If the surface is curved or irregular, this deformation may increase the area of contact, spreading the force per unit area (stress) and decreasing the severity of the resultant injury [17]. If there is an oblique impact where the angle of contact is between 0° and 90°, only a fraction of the kinetic energy is transferred, hence damage will be less than in a normal (i.e., perfectly perpendicular) impact. Mechanical forces that can be exerted include compression, tension, and shear and combinations of such forces. In order to quantify the biomechanics of head injury, the laws of fundamental mechanics and physics need to be evaluated, in addition to the stress and strain limits of the bones and tissues on which they are acting.

Skull fractures

Fractures to the skull occur when a force has been applied in such a way to exceed the strength or the maximum threshold of elasticity of the calvaria. The resulting fracture is determined by the degree of force, the object mass, shape and speed of collision, local anatomy as well as the physiological status of the bone itself, including skull thickness and impact area. This has been summarised in various publications as indicated in Table 1.

Experiments conducted on animal models provide physiological injury data, although scaling laws are necessary to allow translation onto human specimens and are currently imprecise [17]. Not all skull fractures sustained are fatal, and not all head injuries have accompanying fractures. Skull fractures, however, are a useful aid in the interpretation of the point of impact and the nature of the force and object involved. Linear or curvilinear fractures indicate contact of the head by a relatively broad object, often seen in falls or striking with a flat object. More focused impacts, such as those from a hammer, tend to push a small area of bone downward and into brain tissue (a depressed skull fracture). These fractures are comminuted if there are many widely displaced pieces of fractured bone [23].

Falls and blows

A fall involves the head coming into contact with an immovable object such as the ground, and so is similar, in biomechanical terms, to receiving a blow to the head. Similarly, a fall can involve varied forms of associated impact force, be it a fall from one's own height, an accelerated fall from a punch or shove, or a fall from a height. Assessment of resulting injuries from an assault can be quite complex, as the victim can often receive multiple blows from a range of objects or from physical actions such as kicking, punching and stamping. A fall can also occur at any stage in the sequence [23]. Resulting patterns of injury depend on the location and force applied. Previous research into the investigation of applied force is highlighted in Table 2.

It has been suggested that blunt force injury requires greater energy to form a laceration than sharp force injury

Table 1 Summary of previous studies relating to failure of cranial bone

Author	Failure of cranial bone	Models used
Messerer [18]	400–600 kg (M); 300–600 kg (F)	Compression tests—cadaver heads
Nahum [19]	2,450 N (M); 2000 N (F)	Impact mass—cadaver heads
Schneidner and Nahum [20]	~4,000 N	Impact drop tests—cadaver heads
Stalnaker [21]	5,000 N	Pneumatic piston—cadaver heads
Yoganandan [22]	8.8–14.1 kN	Impact—cadaver bodies

Table 2 Forces associated with mechanism of blunt force trauma from the literature.

Author	Associated Force	Associated Mechanism
Allsop [24]	2.3–10 kN Small circular plate	Impact concentrated surface area
	5.8–17 kN rectangular plate	Impact broad surface area
Henn [25]	35 –1,200 N	Kicking and punching
Bohm and Schmidt [26]	500–850 N (M); 350–550 N (F)	Punch
	750–1,200 N (M); 500–750 N (F)	Kicking
Farrugia [27]	634–5,263 N; 2,560 N (average)	Stamp

[28]. The Hat Brim Line Rule (HBL) has been suggested as a means to differentiate between falls and blows, where Ehrlich and Maxeiner [29] and Kremer et al. [30] reported that lacerations of greater than 6 cm were indicative of blows from an object. However, a systematic evaluation of this approach [31] concluded that the use and value of the HBL rule could not be confirmed. Similarly, Kremer et al. [32] noted the need for an improved and adaptable criterion through the analysis of both the location and frequency of lacerations in order to associate a force to the resulting pattern.

In this study we investigate certain mechanisms of blunt force trauma on the basis of the prevalent pattern of injury highlighted through a comprehensive review of autopsied cases coupled with associated literature. The objective was to reconstruct certain mechanisms of single impact injury and measure the associated force to aid pathological differentiation between falls and blows.

Materials and methods

Review of blunt force trauma cases

A retrospective study was conducted involving 377 cases of head injury where an autopsy was undertaken by the Irish State Pathologist's Office from January 2000 to December 2009. Of these, 287 cases included blunt force trauma and were further sub categorised and correlated according to the nature of the injuries encountered. Cases where information was limited, or those involving motor vehicle accidents and gunshot wounds were excluded.

Generation of blunt force injury using porcine models

To associate the mechanics of impact with the pathophysiological changes of the scalp, simulations of single impacts were designed to mimic a fall to the ground, stamping, and blows from two different blunt objects, a hammer and a wooden broom handle.

Morphological and functional data suggests that the skin of the domestic pig is most akin to human skin [33, 34]. Meyer et al. [35] in 1978 noted the possibility of using porcine skin as an experimental model for research into human skin, a suggestion supported by the comparable ratio of dermis to epidermis and epidermal turnover in porcine versus human skin. As such, 5-month-old male and female Landrace pig heads were chosen as suitable models in this work.

A test rig was designed in order to facilitate a perpendicular fronto-parietal impact of the pig head by each implement of interest (hammer, broom handle, training shoe and a piece of wooden flooring). The hammer and broom handle were used to mimic trauma due to a deliberate blow, the training shoe was used to mimic trauma due to a stamping action and the wooden flooring was used to mimic trauma due to a fall.

Each implement was dropped from a fixed height (2.8 m) using a constructed mechanical rig. The weight attached to each implement was modified by increasing the range of applied force. An accelerometer (Silicon Designs Inc. Model 2422) with a frequency response of 3,500 Hz was attached to the drop carriage and provided an output of the maximum voltage on impact. The voltage was converted to acceleration and combined with measurements of the total weight and drop height to generate data on the acceleration versus time, velocity versus time, displacement versus time, force versus displacement and average force of impact using MATLAB® (Version 7.4.0.287 R2007a). This converted the X, Y and Z axes of voltage to acceleration, where only the Z (vertical) was utilized (as X and Y were of much smaller magnitude and due to shock and movement on impact). The specific equations used for the calculations are presented in the Appendix.

Eighteen impact experiments were conducted (each on a different porcine head) for each implement with varying weight in order to generate a range of forces per injury mechanism. Porcine specimens were impacted within 2 to 24 h of slaughter.

Results and discussion

Review of blunt force trauma cases

A total of 287 autopsied cases of cranial blunt force trauma were reviewed. These were classified into two distinct groups; those where the cause of the head injury was known or could be established with some certainty (n=189) and those where the cause was unknown (n=97).

The known group was sub-divided into categories of similar trauma mechanisms. These were falls (from a height, down stairs, or accelerated), blows (blunt objects, stamping, kicking, punching, or axe) and multiple mechanisms, where more than one mechanism was suspected. This approach also facilitated the investigation of the injury pattern associated with each mechanism; results are presented in Table 3 and Fig. 1a and b.

A laceration was present in 44% of cases relating to simple falls, 71% of which were cranial. In cases involving blows from blunt objects a laceration was present in 93% of cases, 89% of which were cranial. Fractures were present in 68% of cases involving simple falls and 75% of cases involving blows from blunt objects In both mechanisms over 90% of the fractures were cranial.

The pattern of injury obtained from examination of cases where the mechanism was known, was applied to the remaining unknown cases, with the aim of identifying the mechanism of injury. However it became apparent that these unknown cases were difficult to discriminate on the basis of injury pattern alone. The use of a database alone was insufficient to determine the mechanism associated with a single impact injury, specifically in the differentiation between falls and blows.

Generation of blunt force injury using porcine models

The blunt force trauma simulations were inflicted on the porcine specimens using a specially designed test rig constructed to deliver a blow of measureable force using the various implements of interest. Under theoretical conditions as outlined in the Appendix, a mass of approximately 3.5 kg dropped from a height of 2.8 m will have an impact velocity of 7.4 m/s and a kinetic energy of 96 Joules. If this energy is fully absorbed so that the impacting implement penetrates to a depth of 0.03 m (3 cm), the deceleration during penetration is approximately 900 m/s^2 (=93 g) and the work-energy principle allows us to infer that the impacting force is 3,200 N.

The weight used was altered in order to vary the range of force applied, for each implement and the forces generated are presented in Tables 4 and 5 and Fig. 2.

Lacerations

Both the hammer and the broom handle caused the greatest incidence of lacerations within the test set. The greater amount of pressure per impact was observed with the focal impact of the hammer compared to all other implements. The pattern of injury observed with impacts from both the hammer and wooden handle were indicative of the implement involved and produced crescent and linear

Table 3 Summary of findings from sub-categories of known head injury cases in relation to orientation and the presence or absence of laceration and/or fracture

Associated Mechanism		Cranial	Facial	Both
	Laceration			
Fall (n=94)	41 (44%)	29 (71%)	14 (34%)	2 (5%)
Fall from height (n=17)	12 (71%)	12 (100%)	2 (17%)	2 (17%)
Fall down stairs (n=13)	5 (38%)	5 (100%)	0 (0%)	0 (0%)
Accelerated fall (n=5)	3 (60%)	2 (67%)	1 (33%)	0 (0%)
Blunt object (n=40)	37 (93%)	33 (89%)	18 (49%)	14 (38%)
Axe (n=6)	6 (100%)	6 (100%)	4 (67%)	4 (67%)
Punch (n=3)	1 (33%)	0 (0%)	1 (100%)	0 (0%)
Kick (n=2)	1 (50%)	0 (0%)	1 (100%)	0 (0%)
Stamp (n=2)	2 (100%)	0 (0%)	2 (100%)	0 (0%)
Multiple (n=7)	7 (100%)	4 (57%)	6 (86%)	3 (43%)
	Fracture			
Fall (n=94)	64 (68%)	58 (91%)	8 (13%)	2 (3%)
Fall from height (n=17)	16 (94%)	16 (100%)	5 (31%)	5 (31%)
Fall down stairs (n=13)	10 (77%)	10 (100%)	1 (10%)	1 (10%)
Accelerated fall (n=5)	4 (80%)	4 (80%)	2 (50%)	2 (50%)
Blunt object (n=40)	30 (75%)	29 (97%)	17 (57%)	16 (53%)
Axe (n=6)	6 (100%)	6 (100%)	4 (67%)	4 (67%)
Punch (n=3)	3 (100%)	1 (33%)	3 (100%)	1 (33%)
Kick (n=2)	1 (50%)	0 (0%)	1 (100%)	0 (0%)
Stamp (n=2)	1 (50%)	0 (0%)	1 (100%)	0 (0%)
Multiple (n=7)	5 (72%)	4 (80%)	2 (50%)	2 (50%)

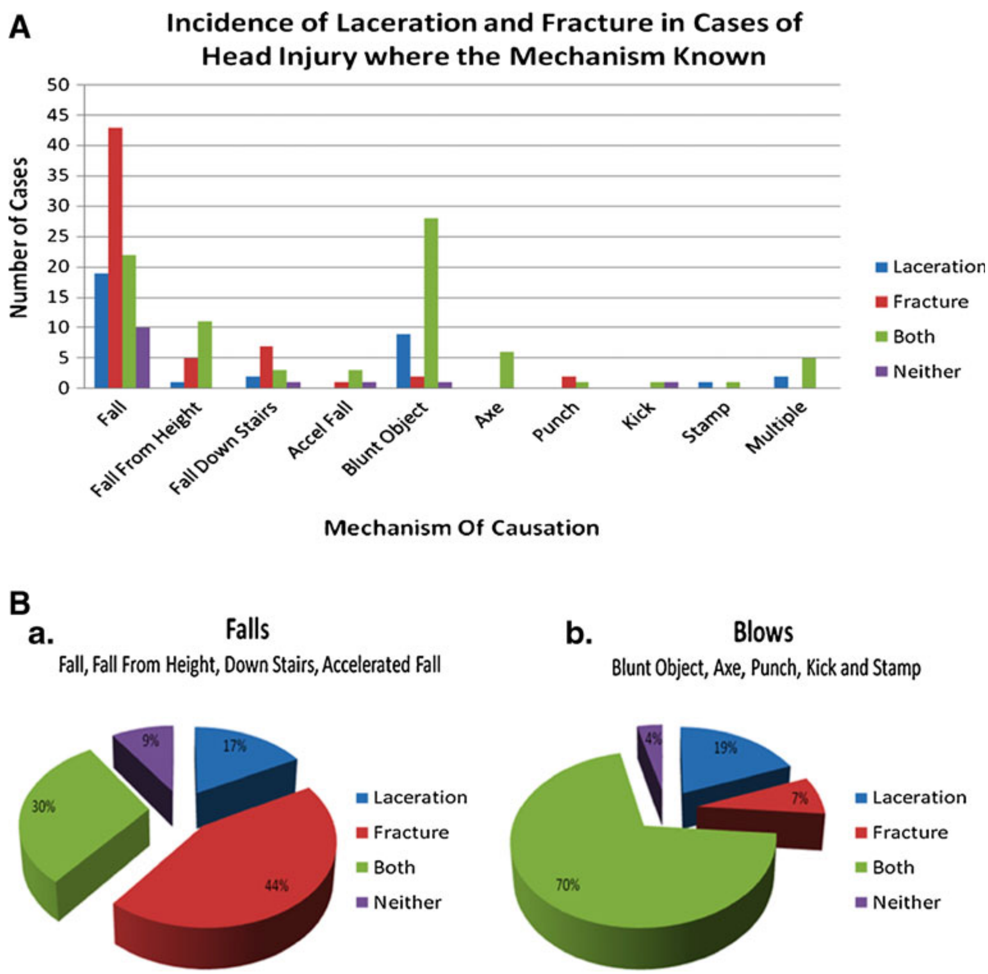


Fig. 1 **a** Bar chart representation of the number of known cases for which the mechanism could be associated with a degree of certainty. The pattern of injury shown involves cases of laceration only, fracture only, both laceration and fracture, and neither laceration nor fracture. **b** Pie chart representations of the pattern of injury associated with known cases involving (i) falls and (ii) blows

shaped lacerations, respectively. Only one laceration was observed from stamping by the training shoe at a force of 5,259 N. The average stamping force has been recorded by Farrugia et al. [36, 37] as 3,500 N. The results also suggest that only falls where forces greater than 4,300 N are experienced resulted in lacerations, though this did not occur in all cases. Observed laceration patterns are presented in Fig. 3.

Overall the results indicate that the minimum force associated with the formation of a laceration is at least 4,000 N. This correlates with the findings of Whittle et al. [38], where the force required to tear human skin was found to range between 2 and 10 kN.

Fractures

The test sets demonstrate a steady increase in the incidence of suture separation with increasing force for the hammer, wooden handle and piece of wooden flooring. Slight skull depression was the second most common skull damage observed between test sets, which would be expected in cases involving more focal impacts over the suture cross section of the fronto-parietal region. The incidence of suture separation occurred slightly less frequently with impacts of the wooden handle, though still increased with increasing force. The hammer, wooden handle and wooden flooring test sets displayed only one linear fracture each.

Table 4 Summary of the implements used, the impact area for the trauma observed and the net weights used to generate the variable forces of impact, at six repeats of each to generate a total of 18 impacts per implement

Implement	Dimensions	Impact area (cm^2)	Net weight (kg)
Hammer	2 cm diameter	3.14	2.67; 3.30; 4.44
Wooden handle	30.5×24 cm	~190	2.84; 3.44; 4.61
Wooden flooring	30 cm (length)×2.2 cm (diameter)	~39.721	3.20; 3.99; 5.13
Training shoe	Women's UK size 5 26 cm (length)×10.5 cm (width) @ ball; 8 cm (width) @ heel	~140	3.22; 3.97; 5.15

Table 5 Summary of actual force and laceration/fracture observations for all tests with each specific implement

Impact	Force (N)	Pressure (kPa)	% Loss in velocity	Laceration observed	Fracture observed
A. Hammer impact repeats					
1.	4,010	1.28	6.01	–	–
2.	4,149	1.32	9.79	Crescent SL	SS and Dp
3.	4,569	1.46	10.91	Double Crescent SL	SS
4.	4,851	1.54	11.74	Crescent SL	L
5.	4,852	1.55	10.68	Crescent SL	–
6.	5,015	1.6	6.91	–	–
7.	5,335	1.7	11.02	Crescent PTL	SS
8.	5,493	1.75	5.48	–	–
9.	5,689	1.82	3.84	Crescent SL	–
10.	5,710	1.82	7.88	–	Dp and SS
11.	6,020	1.92	6.14	–	–
12.	6,350	2.02	6.98	–	Dp and SS
13.	6,712	2.14	9.71	–	–
14.	7,212	2.3	2.82	Crescent SL	–
15.	7,342	2.34	4.75	Crescent SL	–
16.	7,797	2.48	4.96	–	SS
17.	8,632	2.75	5.22	Crescent SL	SS
18.	8,137	2.6	2.96	Crescent SL	Bone loss
B. Wooden handle impact repeats					
1.	4,025	0.1	4.51	–	–
2.	5,333	0.13	4.28	Linear 2 SL	–
3.	5,937	0.15	6.29	–	–
4.	6,176	0.16	3.16	Linear SL	–
5.	6,524	0.16	3.63	Linear SL	SS; L
6.	6,569	0.17	3.63	–	–
7.	6,643	0.17	3.77	Linear SL	SS
8.	7,335	0.18	6.93	–	–
9.	7,442	0.19	2.27	2 Irregular SL	SS
10.	7,505	0.19	2.78	Linear SL	–
11.	7,597	0.19	8.01	–	–
12.	7,885	0.2	3.39	–	–
13.	7,984	0.2	−0.26	–	–
14.	8,261	0.21	9.55	–	SS
15.	8,263	0.21	6.11	Linear SL	SS
16.	8,317	0.21	3.21	–	–
17.	9,460	0.24	1.99	Linear + Irregular SL	–
18.	8,676	0.22	4.89	–	SS
c. Wooden flooring impact repeats					
1.	4,374	0.02	10.67	Irregular SL	–
2.	5,667	0.03	10.14	Irregular SL	–
3.	5,757	0.03	11.24	Irregular SL	SS
4.	6,032	0.03	10.34	–	–
5.	6,041	0.03	8.66	–	–
6.	6,246	0.03	38.22	–	SS
7.	6,255	0.03	10.61	–	SS
8.	6,256	0.03	9.38	–	–
9.	6,797	0.04	9.04	–	–
10.	6,983	0.04	3.31	–	–

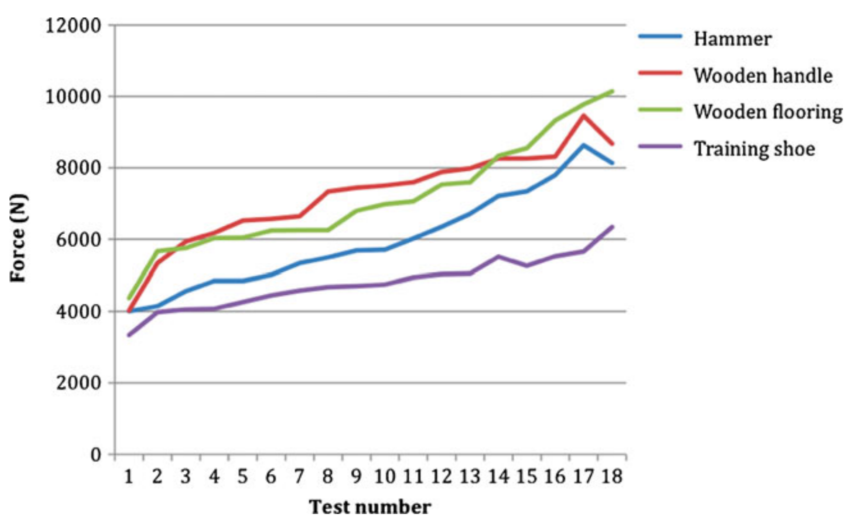
Table 5 (continued)

Impact	Force (N)	Pressure (kPa)	% Loss in velocity	Laceration observed	Fracture observed
11.	7,061	0.04	9.13	–	SS
12.	7,533	0.04	11.41	–	L
13.	7,595	0.04	8.71	–	SS
14.	8,336	0.04	12.70	–	–
15.	8,552	0.05	6.33	Irregular SL	–
16.	9,327	0.05	5.77	Irregular SL	–
17.	9,778	0.05	8.63	–	–
18.	10,140	0.05	8.08	–	SS
d. Training shoe impact repeats					
1.	3,340	0.02	7.40	–	SS
2.	3,981	0.03	8.19	–	SS
3.	4,060	0.03	6.47	–	SS
4.	4,078	0.03	10.86	–	SS
5.	4,265	0.03	14.97	–	SS
6.	4,450	0.03	4.32	–	Dp
7.	4,584	0.03	7.00	–	SS
8.	4,680	0.03	6.44	–	–
9.	4,709	0.03	6.61	–	–
10.	4,753	0.03	6.28	–	Dp
11.	4,946	0.04	4.85	–	SS
12.	5,031	0.04	5.00	–	SS
13.	5,049	0.04	5.88	–	SS
14.	5,511	0.04	5.45	–	–
15.	5,259	0.04	6.95	Irregular SL	–
16.	5,524	0.04	5.96	–	Dp
17.	5,658	0.04	5.82	–	–
18.	6,342	0.05	9.17	–	–

Lacerations classified by increasing severity of superficial (*SL*), partial (*PL*), or full thickness (*FL*). Underlying skull damage classified as suture separation (*SS*), linear (*L*) or depressed (*Dp*)

The '% loss in velocity' is the difference between the theoretical and actual velocity in m/s

Fig. 2 Comparison of the ranges of force generated by each implement



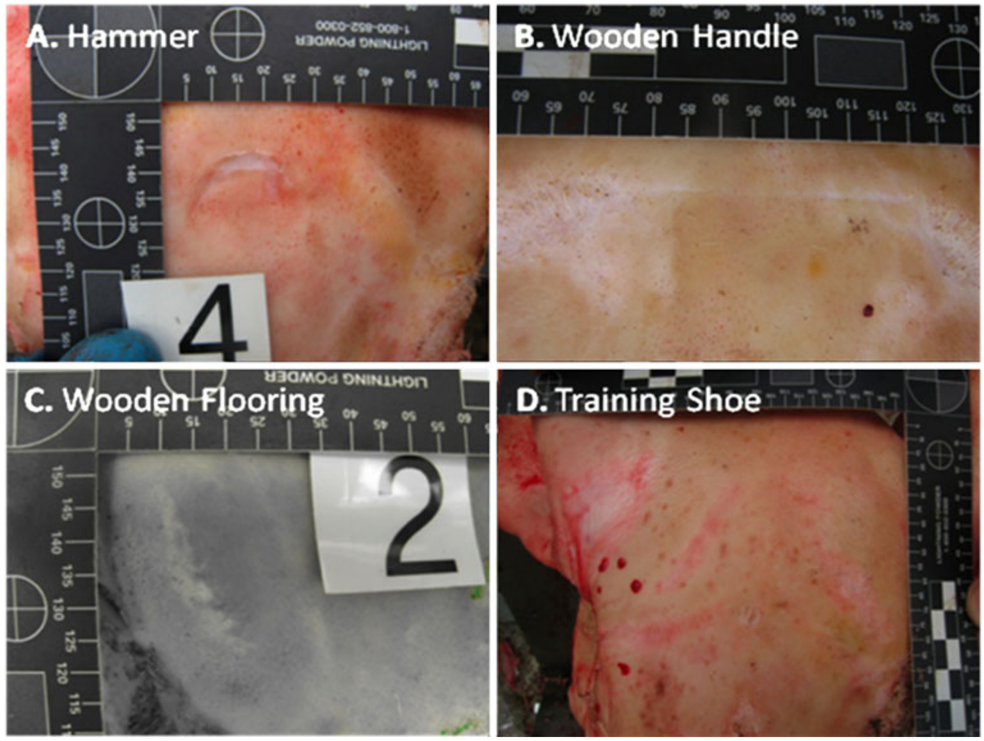


Fig. 3 Lacerations from impact experiments. **a** Crescent-shaped partial laceration, **b** linear superficial laceration, **c** irregular superficial laceration and **d** bruising indicative of trained sole pattern

However, it should be noted that in these cases the skull may have be congenitally or previously weakened prior to experimentation. Impact with the training shoe produced the most suture separations, even at low impact forces.

Combined injury pattern

Twenty-three percent of cases involving simple falls and 70% of cases involving blows from blunt objects presented both laceration and fracture. This suggests that the presence of both injuries increases confidence in differentiating between the mechanism of the injury (fall or blow). This is strongly supported by the experimental findings, where the occurrence of a laceration was more frequently observed in cases where the injury was focused (hammer and wooden handle) and the shape of the laceration indicated the implement used (crescent shaped lacerations and linear lacerations where indicators of impacts involving the hammer and wooden handle, respectively). The sole pattern was also observed on the skin in impacts from the training shoe. Suture separation was the most commonly observed skull damage in all four test sets, particularly with the training shoe and so was not a suitable differential aid. Table 6 illustrates these results.

Conclusions

The occurrence of scalp lacerations and skull fractures are influenced by a range of exogenous and endogenous influences, and as such the majority of cranial blunt force trauma cases require individual assessment. These parameters

Table 6 Observations of the laceration/skull damage obtained

Implement	Observed most common external markers of mechanism for impact experiments	
	Laceration	Skull damage
Hammer	Crescent shaped	Suture separation
Wooden handle	Linear	Suture separation
	Bruising indicating implement shape	
Wooden flooring	Irregular shaped	
	Superficial over thinner subcutaneous tissue	Suture separation
	Broad bruising	
Training shoe	Transfer of sole pattern	Suture separation

include the shape of the bony support, the local thickness of the overlying soft tissues, the impact geometry of the causative implement (striking surface, edge, angle), and the velocity of impact. Notwithstanding this, the pattern of injury associated with the impact experiments, indicate that the occurrence of lacerations and fractures caused by blunt objects are variable and somewhat unpredictable.

Some trends in the data were evident. While there was no distinct correlation between the occurrences of a laceration and increasing force, presumably due to factors associated with the viscoelastic properties of skin, the more focal objects (hammer and wooden handle) produced a greater number of lacerations and where these were produced they provided an indication of the object used. Both the database and experimental findings show that the scalp reacts better to a dispersed force than a localised one, correlating with the anatomical purpose of the scalp to dampen the effect of force via its inherent mechanical properties. Underlying skull damage appears to be more directly linked to the magnitude of the force on impact where all impacts, apart from those involving the training shoe, required a force exceeding 4,000 N for damage to occur. The training shoe, which was delivered in all cases with a force at least equivalent to an average stamping action, produced skull fractures most consistently across the four objects studied. Discrimination between the mechanism of blunt force trauma increased when the occurrence of both a laceration and underlying skull damage was considered. The damage resulting from simulations involving focal impact surfaces (hammer and wooden handle) could be differentiated from those of greater impact surface area (the wooden floor and the training shoe), thus aiding the differentiation between falls and blows.

It should be noted that the review findings are biased towards autopsied cases, and so for a more complete representation of the pattern of head injury associated with specific mechanisms, one might also consider the addition of hospital admittance records. The fact remains that each suspicious death should be investigated individually, combining all possible contributing factors, be it intelligence gathered from scene examinations, additional injuries, or simulated experimentation.

Appendix

The acceleration calculations used took various contributing factors into account, including energy and free fall acceleration due to gravity. Studies have shown that air resistance has a negligible effect on impact velocity during freefalls from less than 50 ft [39], although in guided drop weight experiments where the impactor travels along guide rail tracks, friction effects will reduce the actual impact speed below that predicted by theory. Assuming constant acceleration due to gravity and ignoring friction, the velocity at impact is directly related to the fall height by Newton's equation of motion, where the initial velocity (m/s) equals zero.

$$\text{Newton's equations of motion :} \quad \begin{aligned} V &= U + AT \\ V^2 &= 2AH \\ S &= UT + 1/2A^2 \end{aligned}$$

where V denotes velocity at impact (m/s), U is the initial velocity (=0 m/s when falling from rest), A is acceleration (=9.81 m/s^2 when in freefall due to gravity), T is time (s) and S is distance travelled (m).

The principle of conservation of energy allows us to calculate the impact energy transmitted by an impacting implement to a target.

$$\text{Conservation of energy :} \quad \begin{aligned} E &= \mathrm{PE} + \mathrm{KE} \\ \mathrm{PE} &= MAH \\ \mathrm{KE} &= \tfrac{1}{2}MV^2 \end{aligned}$$

where E is the total energy (J), PE is the potential energy (J) (=0 at the point of impact since H=0 m), KE is the kinetic energy (J) and M is mass of impacting implement and drop carriage (kg).

The work-energy principle can be used in cases where an object in motion has been brought to rest, e.g., when an impactor penetrates an object and stops after a certain depth of penetration. Work done by a force is calculated by multiplying the force acting on the body by the distance it has been moved (assuming that the force and displacement are parallel, as in this case). In order to obtain an average estimate of the impact force, the distance travelled after impact is used with the work-energy principle. The total work done by the force during the impact event is equal to the initial kinetic energy immediately prior to impact.

$$\text{Work} - \text{energyprinciple :} \quad W_{\mathrm{net}} = F_{\mathrm{avg}}D = \frac{1}{2}MV^2$$

where W_{net} is the net work energy (J), D is the displacement during impact (m) and F_{avg} is the average impact force (N).

The accelerometer used in the drop weight experiments allowed us to calculate actual acceleration values during impact; these were used instead of theoretical values, which ensured that friction losses were properly considered in our data. The acceleration values allowed us to calculate velocity, energy, and force using the equations outlined above.

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