Development and Validation of a Physical Model to Investigate the Biomechanics of Infant Head Impact

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ABSTRACT

Head injury in childhood is the single most common cause of death or permanent disability from injury. However, despite its frequency and significance, there is little understanding of the response of a child's head to injurious loading. This is a significant limitation when making early diagnoses, informing clinical and/or forensic management or injury prevention strategies. With respect to impact vulnerability, current understanding is predominantly based on a few post-mortem-human-surrogate (PMHS) experiments. Researchers, out of experimental necessity, typically derive acceleration data, currently a key correlate for head impact injury, by calculation. Impact force, measured at an impacted surface or an impacted head, is divided by the head mass, to produce a "global approximation", a single-generalised head response acceleration value. A need exists for a new experimental methodology, which can provide specific regional or localised response data. A surrogate infant head model, was created from high resolution Computer Tomography scans and 3D-printed, using co-polymer materials with properties closely matched to tissue response data and validated against PMHS head impact acceleration data. High-Speed Digital-Image-Correlation optically measured linear and angular velocities and accelerations, strains and strain rates. The "global approximation" was challenged by comparison with regional and local acceleration data. During perpendicular impacts, regional and local accelerations were up to three times greater than the concomitant "global" accelerations. Differential acceleration patterns were very sensitive to impact location. Suture and fontanelle regions demonstrated ten times more strain (103%/s) than bone, resulting in skull deformations similar in magnitude to those observed during child birth, but at much higher rates. Surprisingly, perpendicular impacts produced significantly greater rotational velocities and accelerations, which are closer to current published injury thresholds than expected, seemingly as a result of deformational changes to the complex skull geometry. The methodology has proven a significant new step in characterising and understanding infant head injury mechanics.

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1. Introduction

Head injury in childhood is the single most common cause of death and permanent disability from injury. Approximately 35,000 children are admitted to hospital in England each year with head injuries, of which 19% are younger than a year of age [1]. Many more children will attend the ED with a head injury, accounting for 9% of all childhood attendances [2].

Paediatric head injury cause and effect is poorly understood, which is a significant limitation when making early diagnoses and informing clinical and/or forensic management. Of the many infant head trauma (HT) cases, clinicians are faced with the difficult task of trying to determine the cases that are a result of abusive head trauma (AHT). This determines whether they inform law enforcement and child protection agencies and provide evidence to support any subsequent legal process. A significant limitation is the paucity of response and injury threshold data available for this type of determination, predominantly due to the rare access to child PMHSs.

Few PMHS studies exist [3, 4 & 5] and only Prange *et al,* 2004 [5], produced quantitative data. Thus, the majority of current infant head impact understanding is based on Prange's study, which investigated three infant postmortem human surrogate (PMHS) heads, impacted from two fall heights (0.15m and 0.30m), at an angle perpendicular to a single surface. Prange, out of experimental necessity, had to derive the current key correlate for head impact injury, acceleration by calculation, by dividing the impact force at the impact surface by the head mass. This limited "global approximation" provides only a single generalised head impact response curve, rather than specific regional or localised responses. Thus, current injury prediction strategies are incapable of considering the significant complexities associated with infant head impacts. Overriding limitations in deriving this information include the rarity of access to child PMHSs and the technical complexities of measuring localised head responses. Faced with these restrictions, clinicians and researchers have often looked to models to help inform their opinions. Physical and computational modeling are an attempt to combine known anthropometry and material properties, with justified approximations for aspects which are not known. With respect to physical modelling, often the limits of suitable materials and manufacturing technologies mean that compromises have to be made. Anthropomorphic test devices (ATDs) are often used to act as surrogates for post mortem human surrogate (PMHS) testing when assessing the possibility of an injury being a result of a given incident. Infant and child head ATDs are, however, based on scaling animal and adult response data and since the paediatric head response is poorly characterised, their specific validity is ambiguous. None of the commercially available ATDs [6,7] represent the separate bone structures or the flexible nature of the infant skull, so there is no appropriate test device that properly represents the infant head impact response.

In addition to the commercially available ATDs, researchers have developed physical models for the investigation of dynamic scenarios [8, 9, 10, 11]. Specific to short falls in young children, Prange *et al*, [10] developed a 1.5-month-old dummy. A limitation of the study, however, was that the head of the dummy was represented by a simplified 2.25mm thick homogenous plastic shell. No account was given to the anatomical complexity of the infant skull plates, sutures and fontanelles, such that it would likely have produced a stiffer response and concomitantly higher output values than an actual infant head.

However, Prange *et al* [10] out of experimental necessity, measured the head impact as if it were a 'rigid body', such that the impact force was measured at the point of impact. The impact acceleration values, a correlate for head injury risk, were calculated from the head impact force, measured at the force plate, by dividing the force by the head mass. Since the impact force is measured at one point on the impact surface, this approach is incapable of providing localised area-specific details with respect to head response.

Coats and Margulies [11] developed a more responsive infant head constructed from copolymer plates, connected by silicon rubber, overlayed by latex to represent the bone, suture and scalp characteristics, respectively. Again localised response measurement was not possible, since the focus was on the global translational and rotational head response, measured by a nine accelerometer array and angular velocity transducer placed at the center of the surrogate head. Head response values were derived from a rigid body assumption.

In response, this present study investigates the development and validation of a 3D printed multi-material physical model, to study the biomechanics of infant head injury, with the capability to measure localised area specific metrics, allowing potential correlation with head injury. Such a physical model is highly significant for investigation of both accidental and non-accidental infant head trauma. The material and methods for the development of the physical model and the drop test methodology are provided in Section 2. The results obtained from the drop tests and the physical model's validation are provided in Section 3, the discussion of the results is in Section 4 and the salient points are specified in the conclusions in Section 5.

2. Materials and Methods

2.1 Design and manufacture of the physical head model

2.1.1 Segmentation of the skull model

To take advantage of improved 3D technology, high resolution skull and brain post mortem imaging was performed on a 10-day old infant for whom no cause of death was found. Post mortem computer tomography (PMCT) imaging was performed with a 64-slice multidetector system (Siemens SOMATOM Definition; Siemens Healthcare, Erlangen, Germany). Volumetric brain PMCT imaging was performed at 120 kV with variable mAs, a pitch of 1, and 0.625 mm collimation. Images were reconstructed with bone algorithms to provide 1.25 mm slices. These images provide the highest resolution acquisition currently obtainable using conventional clinical scanners, although the dose may have been higher than currently in clinical use. Post mortem imaging clearly does not encounter the movement artefacts caused by patient movement, cardiac or respiratory variation and can be conducted until the scans are optimised.

The scans were processed using Mimics Software (Materialise; Leuven, Belgium) to separate the individual bones of the infant head. Separation was achieved by careful use of greyscale 'thresholding' and Mimics tools, together with some manual editing in irregular, geometrically-complex areas. Having separated the parietal (left and right) and occipital bones, the base of the skull and frontal bones were treated as a group, referred to as 'frontal bones' hereafter. Having separated the bones, each one was minimally smoothed to improve its definition as a 3D model, and to reduce its complexity to assist with further computational operations. The occipital bone was significantly recessed in the scan, a movement that occurs during head moulding at birth and typically reverses during the first few weeks of life. In an effort to make the skull more 'generic', that is, representing an infant shortly after birth and to create an easily definable position, the occipital bone was moved outwards slightly to line up with the edges of the parietal bones on each side. Many different 3-Matic tools were required to separate the sutures from the bones and other soft tissues, since their pixel greyscale values in the CT scans were so close to that of the surrounding tissues. Figure (1) shows the final result of the segregation and occipital movement, with a 3D representation of major tissues within the head including cranial bones, skull base, sutures and fontanelles.

2.1.2 Bidirectional properties of infant bone

McPherson and Kriewall [12] determined that infant skull bone is approximately four times stiffer in line with its radial trabeculae, identified as having a fibrous appearance, compared with the perpendicular orientation. To improve the biofidelity of the model bones during impact, an attempt was made to mimic their anisotropic properties. This anisotropic nature could not be directly replicated by the printed materials, however, a solution to this problem was instigated during the development of the 3D data. To mimic the anisotropy a material was selected, which reproduced stiffness in line with the radial direction and very fine radial grooves, emanating from estimated centres of ossification, were designed into the model. The depth of these grooves was determined, such that the second moment of inertia, *I* value, for the circumferential direction was a ¼th of that for the radial, using the equation:

$$I = \frac{bd^3}{12} \qquad \text{..........................} \quad (1)$$

Where the breadth, *b*, was taken as constant and the original thickness of the bone, *d*, was determined as the mean of the thickness, taken at a number of different points. The representation of anisotropic properties for cranial bones is shown in Figure (2).

2.1.3 Manufacturing of the 3D printed physical model.

The physical infant head model was manufactured using Polyjet 3D printing technology, since it allowed the production of a model with more than one material and therefore, more than one material property. During printing, two different materials with different material properties were combined in different proportions to produce different physical properties at different points in the model. The suture geometry and thickness can be precisely controlled and the joints between suture and bone are as strong as any other part of the model. The final 3D CAD design of the head was imported to the 3D printer to produce the physical head model of a 10-day-old child, shown in Figure (3 a).

2.2 Material Selection

2.2.1 Bone and Suture

Bone and suture properties were matched to the few infant response values available in the literature, including the variation for different bones in the head [12, 13]. With regard to the digital materials, whilst the Polyjet technology allows a wide range of material properties to be printed, there is very little numerical data available that defines their specific material properties. To provide strain rate relevant responses, for the purpose of simulating head impact, a number of different commercially available digital materials were tested at low rates in tension (1mm and 300mm/min), and comparisons made with bone samples to find materials that most closely matched published stiffness values for infant cranial bone and suture [12, 13]. Results from the low rate tests, together with the published infant skull and suture values are shown in table (1). The infant skull, comprising the cranial bones and sutures were printed in polypropylene polymers using Stratsys RGD835 Vero White Plus for the frontal and parietal bones, RGD8510 DM Rigid Light Grey 25 for the occipital bone and FLX9870 DM for the sutures

Head structures	Elastic Modulus (MPa)	Matched materials	Polymer response Elastic Modulus (MPa) from low strain rate tests x and y mm/min

Parietal and Frontal Bones	1800 [12,13]	Vero White Plus	2300
Occipital Bone	1200 [13]	Rigid Light Grey	1400
Sutures and Fontanelles	8.1 [13]	Tango Black Plus	7.2
Scalp	1.4 [15]	Latex rubber and polyamide micro fleece	
Brain	0.0272 [14]	Gelatin (10% gelatin: 90% water)	

Table 1. Material properties of 3D printed head structures.

2.2.2 Brain and Scalp

To model the brain, a surrogate brain material, gelatin (10% gelatin: 90% water) was injected into a rubber balloon in the brain cavity through the foramen magnum of the 3D printed head model, in accordance with a previous study [14]. Once the liquid gelatin had set, the foramen was sealed. The scalp model was based on latex rubber, in combination with a polyamide micro fleece, which was found to closely mimic both the elastic and frictional properties of human skin in both wet and dry conditions, respectively [15].

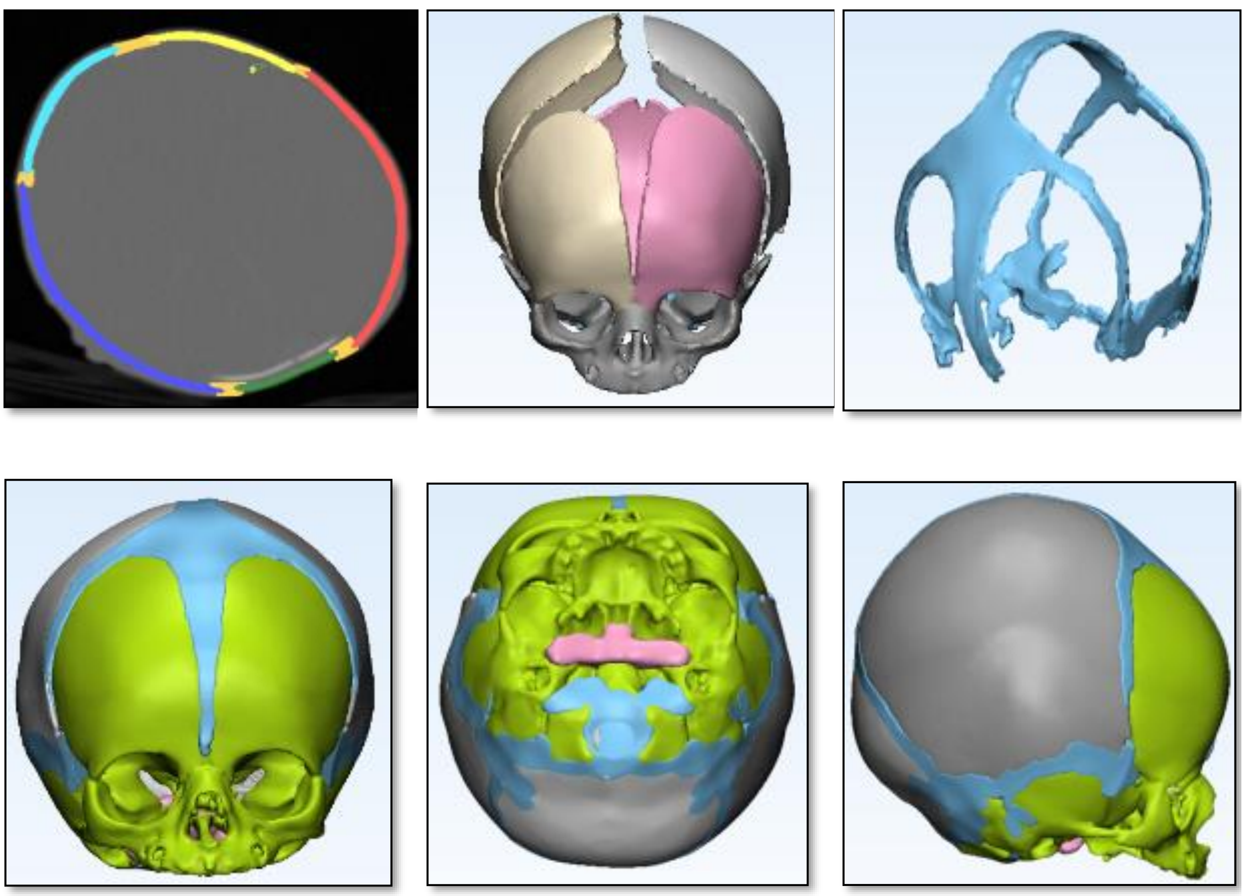
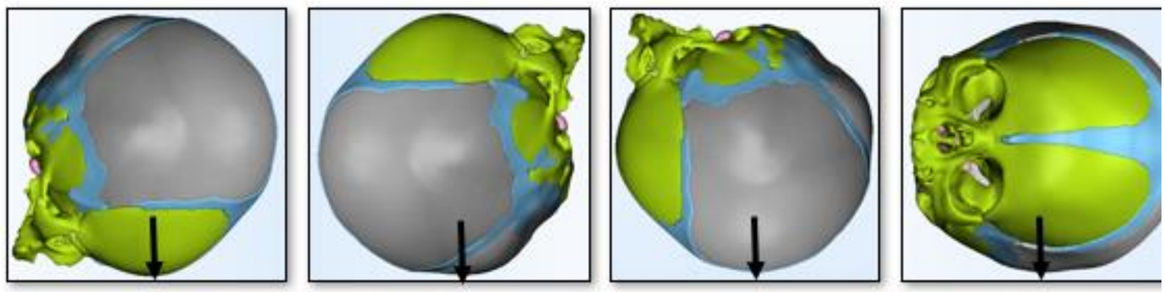


Figure 1. 3D representation of major tissues within the head.



(a) Frontal bone impact (b) Occipital bone impact (c)Vertex impact (d)Parietal bone impact

Figure 2 (a-d). Head drop test orientation.

2.3 Drop tests

For paediatric populations, there are very few experimental tests that can be applied to validate a new physical model of a newborn. Prange *et al*. [5] have conducted infant PMHS head drop test measurements, from two heights (0.15m and 0.30m), with three heads (1, 3 and 11 days old), onto five different impact locations (occipital bone, vertex, frontal bone, right and left parietal bones) onto a steel force plate. The impact accelerations were derived from the force data.

An experimental methodology was established to replicate Prange *et al's* force plate experiments, as shown in Figure (3b). In addition, a High Speed-Digital Image Correlation (HS-DIC) system was applied; high speed video was coupled with an image correlation software to determine how different regions of the skull deformed when subjected to a short fall impact loading. The physical head model was suspended at the required heights and dropped onto the force plate, to permit comparison/validation with Prange's force plate study [5]. The HS-DIC system, which uses two high speed cameras capable of taking images up to 100 kHz, was positioned on the floor to video the impact of the drop tests at 10kHz.

Since the HS-DIC system is typically used to analyse strain in structures when subjected to rapid loading and since the physical model structure was falling through larger distances than is usual when using this technology, interest was in both macro motion of the falling head and the relative micro motion of individual structures. Hand painted speckle patterns were found to be the most effective method for generating a random speckle pattern, since they allowed the identification of their relative positions and calculation of the different physical parameters during their movement relative to one another.

A series of known patterns were used to calibrate the camera, such that the software could distinguish the relative movement of each unique speckle to determine the deformation, velocity, strain, strain-rate and relative rotation between speckles. Thus, a more accurate consideration of the paediatric head model and the impact response to short falls as a whole could be achieved. From the initial trials and set up of the HS-DIC system, the greater the contrast between the speckle pattern and the background, the better the DIC measurements. Thus, it appeared that the brighter the model the greater the contrast between the black speckles and the white background. In Figure (3a), the manually painted acrylic speckle pattern has been added to half of the 3D printed model. For clear contrast, the black paint was used on the white surface of the bones, whilst white was used on the black surface of the sutures.

Out of experimental necessity, the approach taken by Prange *et al* was to consider the head as a rigid body, rather than a soft pliant structure, such that to derive impact acceleration values (a correlate for head injury risk) the head impact force at the force plate was divided by the mass of the head. Since the infant skull is not a rigid body, consisting of a number of flexible plates, flexible sutures and fontanelles, the acceleration-time waveform provides only a “form of average” of total response of the head. This approach is incapable of providing localized area specific details with respect to head response. The HS-DIC system allows for collection of a large amount of information; thus, it was important to determine which points to choose during analysis. Since each physical model impact was different, the selection of the point of analysis had to be made on an individual point-by-point basis, to determine which areas represented the greatest velocity, strain rate and deformation, as shown in Figure (4).

Figure 3(a). 3D printed physical model.

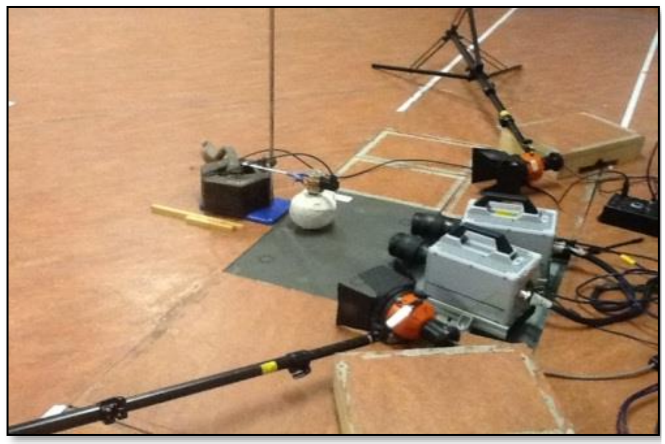
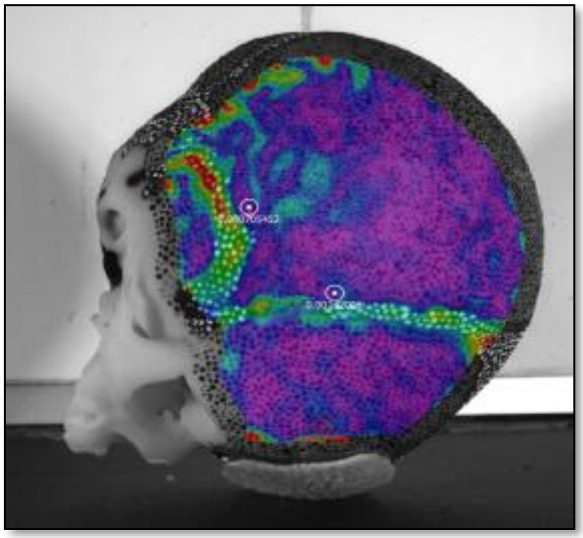
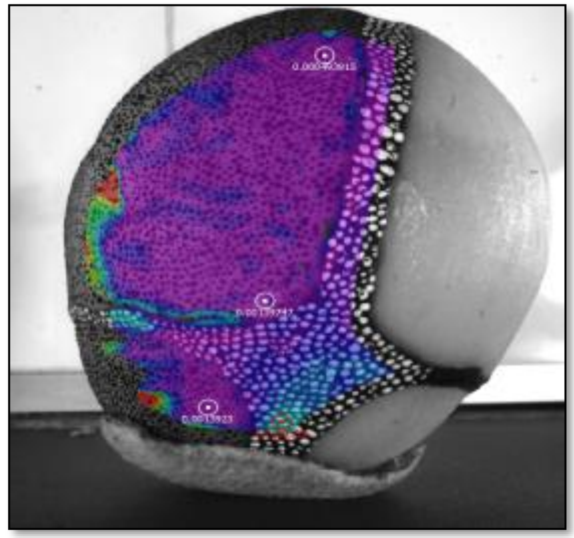


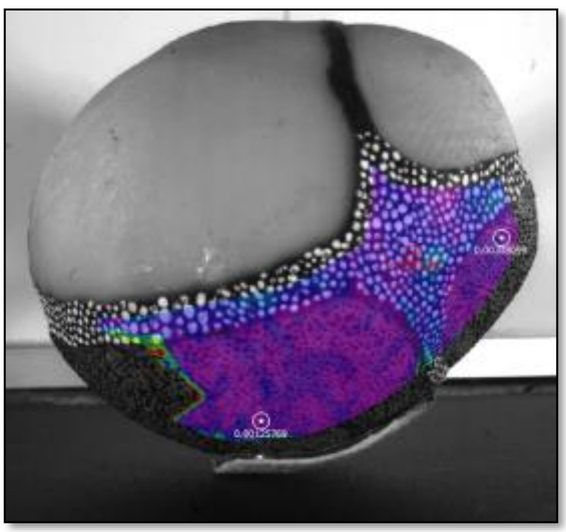
Figure 3(b). Experimental setup of physical model for perpendicular drop onto force plate.



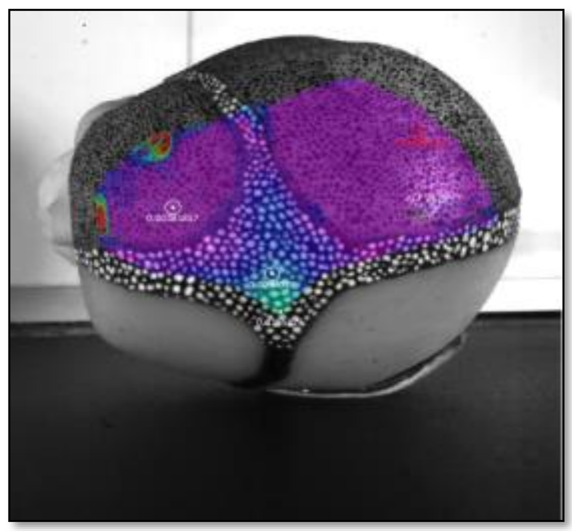
(a) Forehead impact test



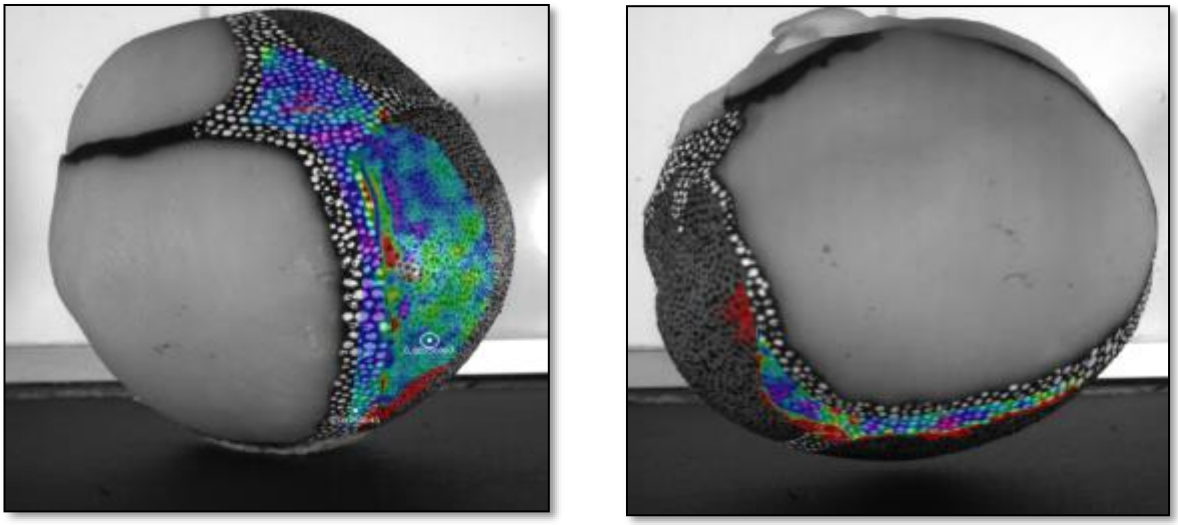
(b) Forehead impact test



(c) Parietal impact test



(d) Parietal impact test



(e) Occipital impact test (f) Vertex impact test

Figure (4 a-f) Digital Image Correlation of the physical model impact tests.

3. Results

3.1 Validation test

Validation of the physical model was achieved by comparing peak impact accelerations against Prange's PMHS model [5]. From Prange *et al*, the mass of the physical head model of the 10-day-old infant was closest to the mass of the 3-day-old infant PMHS head (0.46kg), since, the scalp of the physical model was present only at the impact site, due to the painted random speckle pattern on the bones needing to be visible to the HS-DIC cameras.

From Figure 5, it can be seen that the physical model test and Prange *et al's* peak resultant accelerations for all the 8 drop tests show a close correlation for all conditions, except for the occipital bone, at 0.15m and 0.30m and the forehead bones at 0.30m. The average peak accelerations for the occipital bone (82.1g and 116.7g) and forehead (117.6g) were significantly greater than average peak acceleration values from Prange, which were 46g, 72.1g and 82.1 g, respectively.

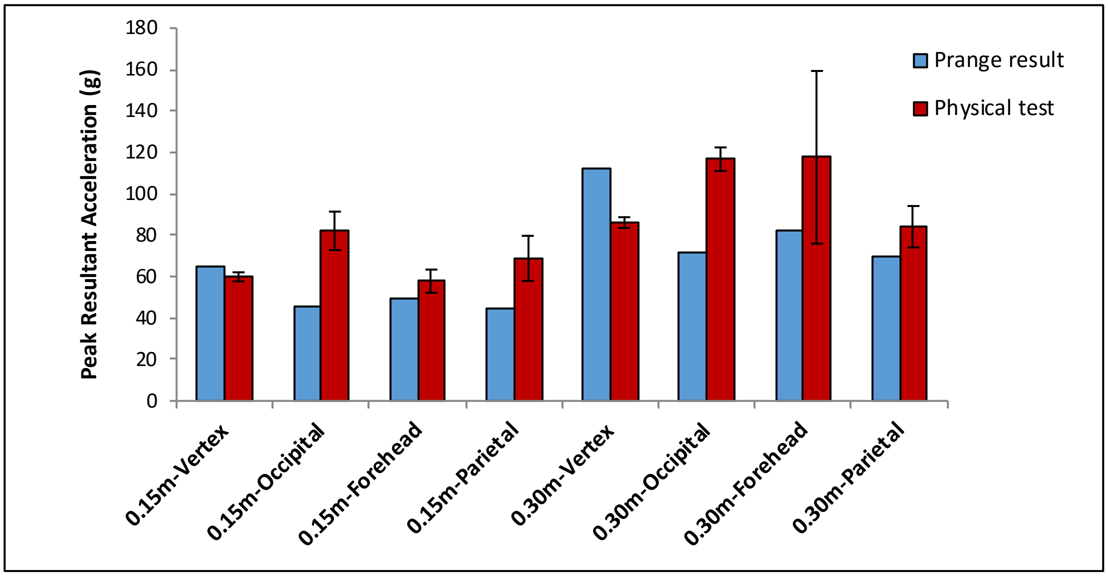
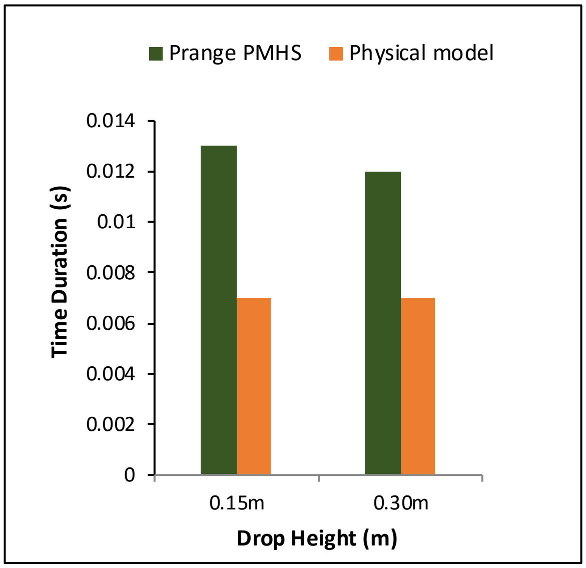


Figure 5. Peak acceleration and drop height for infant PMHS and physical model impact tests.

During the validation process, the impact time durations between the PMHS tests and physical model were compared. From Figure 6, the average time durations, for the physical model during both 0.15m and 0.30m drop height impacts, are observed to be slightly shorter than those in Prange's tests. It is of particular interest that during different drop tests, the peak accelerations appeared to be close to the Prange tests (Figures 5 and 7), whilst the impact time durations were not (Figures 6 and 7). The acceleration-time contact curve for the forehead area, from a 0.30m height, shown in Figure 7, can be used to evidence this point. The peak impact acceleration, from the physical head model, is closer to the peak impact acceleration measured by Prange, while the impulse of the physical model force time response is 3.4 N*s, compared to 4.4 N*s from the PMHS.



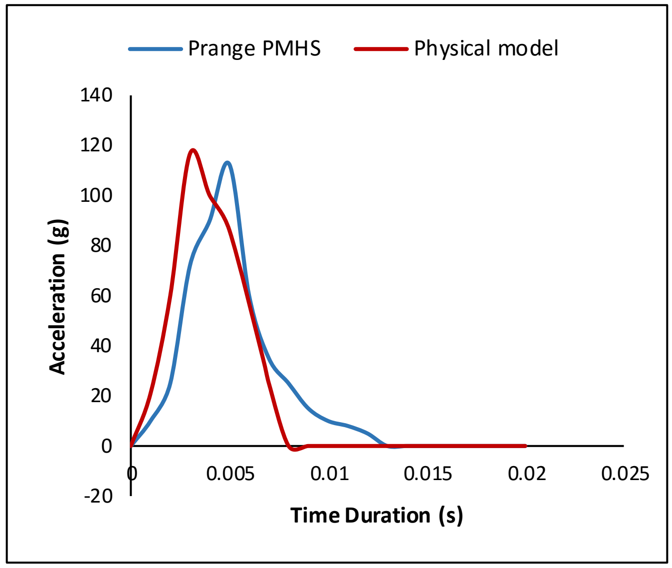


Figure 6. Comparison of impact time duration between PMHS test and physical model.

Figure 7. Acceleration-time contact curve for the forehead area from a 0.30m height drop.

3.2 HS-DIC system results

3.2.1 Translational and rotational accelerations

The physical head model was validated for acceleration-vs-time, against published PMHS data by the same drop test method using a rigid body assumption. To assess this assumption, HS-DIC was used to measure linear and angular acceleration response from different impact locations at two low drop heights, compared with the impact response of the physical model from the force plate tests.

It appeared, from Figure 8, the translational acceleration from the DIC tests are significantly higher than those from the force plate response. Interestingly, the greatest percentage differences are in frontal impacts, where at 0.15m, the mean highest recorded bone acceleration was 257% greater than the mean value for the force plate. The smallest increase, from the rigid body assumption to the DIC bone result, was for the 0.15m drop onto the vertex, where the difference was 30% greater.

Figure 9, demonstrate that the physical model angular acceleration responses were different between different impact locations and different drop heights. The greatest peak angular acceleration, for each drop test, was measured. Peak angular accelerations for parietal views of vertex impacts were as high as 4121 rad/sec^2, compared to the frontal and occipital views, which were 1779 and 1107 rad/sec^2, respectively.

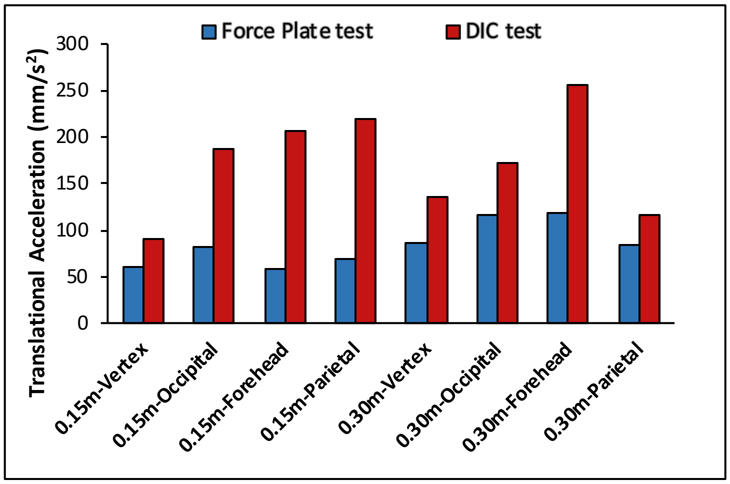


Figure 8. Physical model comparison of translational acceleration from the force plate and DIC system.

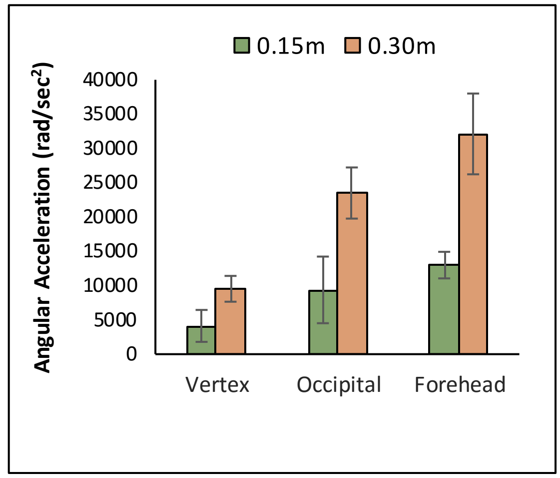


Figure 9. Physical model maximum angular acceleration from different impact locations at two drop heights.

3.2.2. Strain and strain rate.

The HS-DIC system was applied to determine which areas of the physical model produce the greatest deformation and strain rate. It was observed that the majority of the strain and strain rate was dissipated by the suture and fontanelle areas. Figures 10 and 11 illustrate the maximum strain and strain rate values in the sutures and fontanelle region, where the head was dropped from two heights onto different head impact locations. The suture and fontanelle areas produce strain values between 1000%/s and 2000%/s.

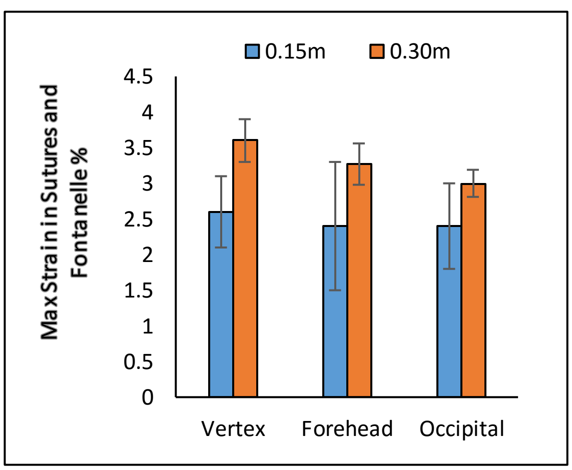


Figure 10. Maximum strain and drop height from different impact locations of physical model.

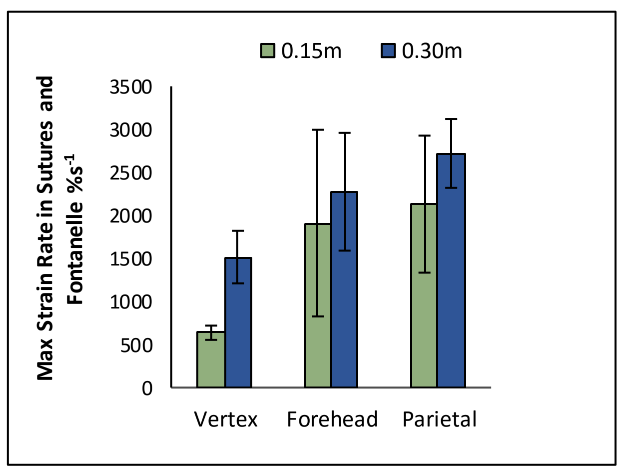


Figure 11. Maximum strain rate and drop height from different impact locations of physical model.

4. Discussion

A geometrically accurate physical head model was produced from high resolution human infant CT images using materials whose responses were matched to published infant human tissue data, which was printed by a Polyjet 3D printer. The head model was validated against infant PMHS head impact data, as shown in Figure (5). Peak resultant acceleration for all the impact test cases demonstrate a close relationship to Prange *et al's* PHMS impact tests, excluding the occipital and forehead test.

For vertex impacts (Figure 2c), the head peak acceleration response from the force plate was very close to the Prange study at 0.15m, with a 7% difference, compared to the 0.30m impact height that produced a 23% difference.

For the occipital bone impacts (Figure 2b), the description of the impact location by Prange *et al*, was “occipital” only. Due to the complex geometry of the infant skull, this meant that there were at least two possible impact locations. For the purpose of experimental transparency, the center of the occipital bone was considered as the impact point, close to the posterior fontanel and parietal bones as shown in the HS-DIC images (Figure 4e). It is not clear what specific orientations were tested by Prange, but it is noteworthy that the PMHS results are closer to the physical response at 0.15m than at 0.30m.

Perhaps of greater significance, is the position of the occipital bone in relation to the parietal bones. The occipital bone of the physical model was moved to line up with the edges of the parietal bone, corresponding to a position that might be expected in an infant some days/weeks after birth, since the depressed occipital bone is a feature of head moulding during birth. The bone moves out of line relative to the parietal bones during birth, though can remain displaced throughout the first few weeks of life, especially if an infant is placed on its back, to sleep. It is quite possible that the PMHS heads used by Prange had their occipital bones depressed, this could have resulted in potentially less impact loading being transmitted to the surrounding parietal bones, such that the bone’s response relies more on the support of the brain and pressure inside the head. It would seem possible that this could mean a less stiff response and hence, lower acceleration, were this to be the case. This may well explain the higher accelerations observed in the physical model as shown in Figure 5.

For parietal impacts (Figure 2d), the results in Figure 5 appeared to be affected by minor perturbations in impact location, combined with the position of the occipital bone. With the occipital bone in line with the parietal bone, the head’s structure, from a parietal loading perspective, is likely to be stiffer at the posterior side. This is most likely if the impact is slightly towards the posterior of the parietal bone, rather than the anterior and could explain the higher accelerations shown in Figure 5, since the model’s occipital bone was in line, compared to the PMHS bones, which may have been depressed.

With regard to the forehead drop test (Figure 2a), by analysing the HS-DIC images (Figure 4b) during the point of impact, variations in deformational response were observed. These variations could be explained by slight alterations in the drop test orientation. Whilst any drops that were seen to strike the facial bones before the forehead were excluded, since the facial area is relatively stiff and buttressed by the many complex bone structures, the impacts onto the lower part of the forehead produced significantly higher accelerations. However, when the impact occured at the upper part of the forehead, lower acceleration profiles were observed as the loads were being transferred to the more flexible parietal bones and anterior fontanelle.

The peak accelerations for most impact locations from the physical test are slightly higher than Prange's, however, the values still fall within Prange's published response corridor. Thus indicating, that the stiffness of the physical head model is greater than the PMHS head. It appears, from Figure 5, that average peak accelerations for the physical head model are significantly different for different drop heights, but do not vary between impact locations, which is in good agreement with the conclusion from Prange's experimental tests. When comparing the results and taking into consideration the inherent variance that exists between different specimens, it can be considered that the responses from the physical paediatric head model and PMHS head are in favorable agreement under the experimental drop conditions [5].

The peak impact acceleration measured by PMHS test is closer to the peak impact acceleration from the physical model as shown in figure 7. Where the physical model's head impact curve has a bell shape, which closely resembles the shape of the PMHS curve for the same impact height of 0.30m and same impact area the forehead. However, there is a difference in impact time duration between PMHS test and the physical model for the same impact height and location Figure 7. The variations in impact time duration were significant at different drop heights, shown in Figure 6, and are likely due to the differences in stiffness between PMHS and the physical model.

Following the validation of the physical head model for acceleration-vs-time, against published PMHS data using the force plate and rigid body assumption, the HS-DIC system was used to measure the linear and angular accelerations and strains to assess the rigid body assumption, by comparing global, regional and local accelerations.

The DIC system allowed the elucidation of the rigid body approximations that have been reported in the published research to date. The translational accelerations, measured at discrete sites on the skull as shown in Figure 8, appeared to be significantly higher than those measured at the force plate. This is likely a result of the mass of the skull bones constituting a comparatively small measure of the overall mass of the head. Thus, the skull bones are likely to come to rest more quickly than the more massive brain structures, quantification of the relative accelerations will prove invaluable in any subsequent modeling of the cause of meningeal haemorrhaging.

The DIC system was used to measure local area deformation at the outside of the skull rather than considering the head as a lump mass viewed from the perspective of acceleration against time as shown in Figure 4. The speckle DIC system provided the kinematic data associated with a low height fall through 0.15m and 0.30m.

Angular acceleration, shown in Figure 9, confirms that the infant head is very deformable and when it is deformed, it will produce different angular acceleration responses in different structures of the head at different times. This is extremely valuable for quantifying deformation of the head for determination of both the nature and the risk of a head impact. The DIC software concomitantly measured the angular acceleration, at points across the head and allowed comparison between regions, such that the greatest peak angular acceleration for each impact could be quantified.

From the DIC system, shown in Figure 10 and 11, it appears that during impact the majority of the strain was dissipated by the sutures and fontanelle areas of the head. Given that the infant head naturally deforms in these areas during the quasi-static loading of child birth, this response might be expected.

During impacts, however, the head is loaded at a much higher rate, such that the skull is observed to deform and as a consequence, the brain and other structures inside the cranium deform at a higher rate. The DIC results for the strains within the bones were very much more “noisy” than the sutures and fontanelles, but do suggest an engineering strain order of magnitude of approximately 0.3%, for both 0.15m and 0.30m drops. By contrast, due to their much larger strain and rate values, the results for suture and fontanelle areas are “cleaner”, with magnitudes of strain ten times (10x) greater, between 1000%/s and 2000%/s. These significant skull deformations that are similar in magnitude to those observed during a natural child birth [16, 17], but at much higher rates Although there is still no clear difference between 0.15m and 0.30m drops. Whist it must be acknowledged that the model represents only one age matched infant, who has been assumed to be representative of both Prange *et al's* PMHS and further, of infants in general, there will certainly be small variations between individuals, with respect to skull composition and geometry, it is, however, anticipated that the principles of the observations will hold true.

5. Conclusions

A geometrically accurate physical head model was produced from high resolution human infant CT images using materials whose responses were matched to published infant human tissue data, and printed by a Polyjet 3D printer. The head model was validated against infant PMHS head impact data. Subjecting the head model to impact tests, demonstrated a close correlation with PMHS head test data. Comparing the physical model with the results of the PMHS drop tests, demonstrated that using a rigid body approximation to determine acceleration, has significant shortcomings for the purpose of determining head response, due to the flexible nature of the infant skull. High Speed Digital Image Correlation was shown to be highly effective in establishing acceleration data. Differential acceleration patterns were observed to be very sensitive to impact location, corresponding to epidemiology findings, that falls from similar heights onto different skull locations result in very varied clinical outcomes. Surprisingly, significant rotational velocities and accelerations were observed, which are closer to current published injury thresholds than expected, seemingly as a result of deformational changes within the complex skull geometry. The suture and fontanelle regions demonstrated ten times more strain (103%/s) than the bone, resulting in skull deformations that are similar in magnitude to those observed during a natural child birth, but at much higher rates. Strain rates in sutures and fontanelles, for impacts from 0.15m and 0.30m, were of the order of 10^3%/s. Accelerations at certain areas of the skull structure were observed to be two to three times greater than concomitant global acceleration values and significant variation was demonstrated across the surface of the skull. Strains in the bone, suture and fontanelle areas, for impacts from 0.15m and 0.30m were shown to be between 0.3% and 3%, respectively. The methodology of combining high resolution scan data with 3D printing technology, response matched polymeric printed materials and PMHS materials and impact response data has been proved to produce a significant new step in characterising and understanding infant head impact injury mechanics.

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