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Review

An investigation of cranial motion through a review of biomechanically based skull deformation literature

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Abstract

Objectives

There is ongoing debate over the existence of cranial motion resulting from manual manipulation during Cranial Osteopathy (CO). The purpose of this study was to review and summarize the literature regarding cranial mobility and human cranial stiffness in order to evaluate the validity of cranial movement in humans due to manual manipulation.

Methods

In Part I, the literature was reviewed to determine the existence and extent of cranial motion in animals and humans. In Part II, the literature was reviewed to determine the stiffness of the human cranium. In Part III, a biomechanical analysis was performed to determine the amount of force necessary to cause cranial deflections reported in the studies identified in Part I, using published skull stiffness values reported in the studies identified in Part II.

Results

Skull deflection across the cranial sutures of animals ranged from 0 μm to 910 μm. Cranial vault deflection in living humans was reported to range from 0.78 μm to 3.72 μm. Reported human skull stiffness values ranged from 390 N/mm to 6430 N/mm depending on the region of the skull and the method of loading. Based on the range of skull stiffness values, it was determined that an applied force between 0.44 N and 23.2 N would be required to cause 0.78 μm of deflection, and between 2.09 N and 111 N would be required to cause 3.72 μm of deflection.

Conclusion

Externally applied forces and increases in intracranial pressure can result in measurable cranial motion across the cranial sutures in adolescent and adult mammalian animal species, and measurable changes in cranial vault diameter in post-mortem and living adult humans. However, the amount of cranial motion may vary by subject, the region of the head to which forces are applied, and the method of force application. Given that the forces required to generate reported cranial deflections in living humans are within the range of forces likely to be used during CO, it is reasonable that small amounts of cranial deflection can occur as a result of the forces applied to the skull during CO.

Keywords

- Cranium;
- Osteopathy;
- Rhythm;
- Cranial movement;
- Skull deflection;
- Stiffness;

- Biomechanics

Introduction

Cranial Osteopathy (CO), also referred to as Osteopathy in the Cranial Field (OCF), is used by osteopaths to treat a variety of medical conditions and symptoms resulting from injuries. Several studies report that CO has successfully improved symptoms of neurological and medical problems in children[1, 2 and 3] and adults[4 and 5] including but not limited to: infantile colic,[2] infantile gastroesophageal reflux,[3] vision impairment,[4] and lasting effects of brain injury.[5] The potential improvement of symptoms resulting from the lasting effects of brain injury are of particular importance due to the prevalence and morbidity of these injuries in athletics,[6, 7, 8, 9, 10, 11 and 12] automobile collisions,[13, 14, 15 and 16] and the military.[17, 18 and 19]

In 1899, Dr. William G. Sutherland began the development of CO.[20 and 21] Practitioners of CO believe that rhythmic motion of the cranial bones exists due to fluctuations of the cerebrospinal fluid pressure[23] and arterial blood pressure.[24] The fundamental theory behind the cranial techniques used during CO is that passive palpation can be used to detect spontaneous or rhythmic movement of cranial bones, and the application of selective pressure to the cranial bones can manipulate the rhythmic movement of the bones and intracranial fluids to achieve a therapeutic outcome.[20, 21, 22 and 23] Therefore, the belief that the cranial vault is both a mobile and compliant structure is critical to the basis of CO, and without this motion the rationale of therapeutic cranial techniques is unsubstantiated.[23]

The human cranium (or skull) is comprised of eight plate-like bones: the frontal bone, left and right sphenoid bones, left and right parietal bones, left and right temporal bones, and the occipital bone.[25] The areas where these bones join together are known as cranial sutures. During infancy the bones of the skull are not rigidly joined together.[25] A membrane (called a fontanelle or “soft-spot”) made up of strong, fibrous, elastic tissues fills the space where two sutures join.[25] After approximately 12–18 months, the cranial bones begin to grow together and fuse as part of normal development.

In both children and adults, cranial bone mobility resulting from CO has been described to occur mainly at the cranial sutures.[23 and 24] The size and flexibility of the fontanelle in an infant skull creates a non-rigid cranium structure, which likely results in increased cranial mobility in children compared to that of an adult. This could potentially explain the success of CO in treating the symptoms of neurological and medical problems in children.[1, 2 and 3] However, many medical professionals believe that motion across the cranial sutures is not possible in adults due to fusion of the sutures with age.[21 and 23] Some studies have reported that the cranial bones are completely fused between 13 and 19 years of age.[26 and 27] Other studies have reported that the closure of the sutures varies by individual, and that while the sagittal, coronal, and lambdoid sutures close by 40 years of age complete boney union of all sutures is not complete until the 8th decade of life.[28 and 29] Still other studies indicate that a rigid union of sutures never occurs, and therefore some degree of mobility at the sutures is possible throughout life.[30, 31 and 32] The discrepancies in the literature regarding closure of the cranial sutures has lead to an ongoing debate over the existence of cranial mobility in adults, particularly across the sutures.[21 and 23]

Although several studies have been conducted to investigate the existence of cranial bone movement, there have been mixed findings regarding the existence and degree of cranial mobility.[33, 34, 35, 36, 37, 38, 39 and 40] In addition, provided that the cranial vault is a mobile and compliant structure, there is debate as to whether the forces generated from manual manipulation during CO are sufficient enough to cause movement of the cranial bones.[30 and 38] This study approached the debate of cranial mobility as a result of CO from an engineering perspective. The purpose of this study was to review and summarize the literature regarding cranial mobility and human cranial stiffness in order to evaluate the validity of cranial movement in humans due to manual manipulation.

Methods

This study was conducted in three parts. In Part I, the literature was reviewed to determine the existence and extent of cranial motion in animals and humans. In Part II, the literature was reviewed to determine the stiffness of the human cranium. In Part III, a biomechanical analysis was performed to determine the amount of force necessary to cause published cranial deflections (Part I) using published skull stiffness values (Part II).

Part I: review of cranial motion

A review of the literature was performed to identify published studies investigating cranial motion in animals and humans. The literature search was performed using one search vehicle: PubMed. Various combinations of the following search terms were used: cranialsacral, cranial, skull, therapy, manipulation, intracranial, pressure, distraction, coronal, parietal, bone, suture, mobility, movement, deflection. The reference sections of the relevant publications identified through these searches were then searched to identify additional relevant publications. All studies which quantified the cranial motion of animals and humans due to spontaneous motion, externally applied force, or an increase in intracranial pressure were briefly summarized. Measured values of cranial motion were reported for each study.

Part II: review of human skull stiffness

A review of the literature was performed to identify published studies investigating the mechanical stiffness of the whole human skull or regions of the human skull. The literature search was performed using two search vehicles: PubMed and the Society of Automotive Engineering (SAE) Safety Series. Various combinations of the following search terms were used for both search vehicles: biomechanical, mechanical, response, stiffness, human, skull, head, impedance, force/deflection, tolerance. The reference sections of the relevant publications identified through these searches were then searched to identify additional relevant publications. All studies which quantified and reported the mechanical stiffness of the whole human skull or regions of the human skull were then summarized, including a brief description of methods and the reported stiffness values.

Part III: biomechanical analysis

The amount of external force required to achieve the human skull deflections reported in the literature (Part I) was estimated using the average human skull stiffness reported in the literature (Part II). This was done by multiplying the average skull stiffness (N/mm) for a given study by the skull deflections (mm) reported in the literature.

Results

Nine published studies investigating cranial motion are reviewed in Part I of the results. Six studies performed tests on animal specimens, and three studies performed tests on human specimens. Part II reports human skull stiffness values from nine studies and results of the biomechanical analysis regarding cranial motion. Part III reports the estimated amount of external force required to achieve the human skull deflections reported in the literature.

Part I: review of cranial motion

Animal studies

Michael and Retzlaff[33] evaluated the motion of the parietal bone in adult, female monkeys, *Saimrir sciureus*, due to externally applied forces and passive spinal motion. The heads of the primates were rigidly immobilized with a stereotaxic frame during the experiments. Lateral immobilization was accomplished using tapered rods inserted in the external auditory meatus. Vertical immobilization was accomplished using a flat bar pressing upward against the roof of the mount and hook-like bars pressing downward on the infra-orbital ridges. The frequency of parietal bone movement was monitored using a transducer attached to a screw placed in the right parietal bone. Michael and Retzlaff[33] reported that both externally applied forces and passive spinal motion resulted in parietal bone motion. This study noted that one pattern of parietal bone motion was synchronized with respiration rate. A second pattern of motion was also observed, but could not be attributed to heart rate, respiration rate, or venous pressure. Unfortunately, quantitative measurements of cranial motion were not reported in terms micrometers (μm).

Retzlaff et al.[34] performed a follow-up study to evaluate the motion of the parietal bones in adult, female squirrel monkeys. Parietal bone motion was quantified using force displacement transducers attached to screws placed into the approximate midpoint of the left and right parietal bones. The heads were loosely mounted in a stereotaxic frame, as opposed to rigidly immobilizing the head like Michael and Retzlaff.[76] The results of the study showed that there was oscillatory motion of the parietal bones. In addition, Retzlaff et al.[34] noted that the slow and fast oscillatory motion of the parietal bones was directly correlated with respiration rate and cardiac activity, respectively. Although Retzlaff et al.[34] provided example data plots of these measurements, the scale was not indicated. Therefore, quantitative measurements of cranial motion could not be determined in terms of micrometers (μm).

Oudhof and van Doorenmaalen[35] used strain gauges to measure deflection between the sagittal and coronal sutures in six-week old and three-year old beagles. Deflection of 5–10 μm was observed in the beagle puppies (Table 1). In addition, it was found that the skull deflection patterns matched changes in aortic pressure and an electrocardiogram (ECG). While no deflection was seen in the skulls of the adult beagles, Oudhof and van Doorenmaalen[35] noted that a lack of measurable movement in adult skulls could be due to the insufficient instrumentation sensitivity.

Table 1. Cranial movement in various animal specimens.

Author	Specimen	Number of Subjects	Spontaneous Motion vs. Applied Force	Observed Skull Deflection (Yes/No)	Skull Deflection (μm)
Michael and Retzlaff[33]	Adult Primate	N/R	External Compressive Force & Spinal Motion	Yes	N/D/R
Ratzlaff et al.[34]	Adult Primate	N/R	Spontaneous Motion	Yes	N/D/R
Oudhof and van Doorenmaalen[35]	Beagle Puppy	N/R	Spontaneous Motion	Yes	5–10
	Adult Beagle	N/R	Spontaneous Motion	No	0
Pitlyk et al.[36]	Dog	6	Increased Intracranial Pressure	Yes	N/D/R
Adams et al.[37]	Adult Cats	N/R	Internal and External Forces	Yes	17–75
Downey et al.[38]	Rabbit	13	Tension Across Suture	Yes	310–910

Note: N/R is Not Reported; N/D/R is No Deflection Values Reported.

Table options

Pitlyk et al.[36] used strain gauges to measure cranial movement due to changes in intracranial pressure in six dogs. Intracranial pressure was increased with a balloon attached to the end of a catheter and inserted into the subarachnoid space of the brain or by injections of saline into the spine. Strain gauges were attached to an apparatus that arched mediolaterally across the head to measure changes in skull diameter. Unfortunately, the quantitative measurements of cranial motion were only reported in terms of the millivolt (mV) output of the strain gauges, as opposed to micrometers (μm). Although deflection data for cranial motion was not reported in the publication, the study concluded that cranial movement correlates to an increase in intracranial pressure (Table 1).

Adams et al.[37] measured movement across the sagittal suture of anesthetized adult cats. An apparatus with two strain gauges to measure cranial motion was placed across the sagittal suture. One gauges measured deflection across the sagittal suture, while the other gauges measured the degree of parietal bone rotation. Cranial motion was induced by various external and internal forces exerted on the skull.

External forces included manual compression of the sides of the skull, manually applied downward force on the sagittal suture, and a 2.2 N compressive or tensile force applied by a spring across the parietal bones. Three different methods were used to exert internal forces on the skull which aimed to increase intracranial pressure: induced hypercapnia by inhaling excess CO_2; intravenous injection of norepinephrine; and injection of artificial cerebrospinal fluid (CSF). Adams et al.[37] reported that measurable motion across the sagittal suture occurred from all externally and internally applied forces on the skull, and ranged from approximately 17–75 μm depending on the method of force application and animal (Table 1 and Table 2). Adams et al.[37] noted that application of force to the sides of the head caused the parietal bones to move closer together and rotate inward, which resulted in an increase in intracranial pressure as well as transient changes in heart and respiratory rate. After removal of the applied force, the position of the parietal bones, heart rate, and respiratory rate returned close to normal. Conversely, the application of a downward force on the sagittal suture caused the parietal bones to move further apart and rotate outward, but did not result in any changes in intracranial pressure, heart rate, or respiratory rate. The 2.2 N compressive force applied by a spring caused the parietal bones to move closer together, while the 2.2 N tensile force applied by a spring caused the parietal bones to move further apart.

Table 2. Cranial movement in anesthetized cats reported by Adams et al.[37]

Internal or External Force Application Method		Skull Deflection (μm)
External Force	Manual compression of sides of head	−65
	Manual downward force on sagittal suture	30
	Spring compression across sagittal suture	−40
	Spring tension across sagittal suture	75
Internal Force	Induced hypercapnia	30
	Norepinephrine injection	−50
	Artificial Cerebrospinal Fluid (CSF)	17–70

Note: Positive deflections indicate separation of the parietal bones; Negative deflections indicate compression of the parietal bones.

Table options

A study conducted by Downey et al.[38] observed separation across the coronal suture in New Zealand white rabbits. Brackets were mounted to the skull on either side of the suture and used to apply tension across the suture. All subjects were exposed to 5, 10, 15, and 20 g of force. One subject experienced additional loads of 100 g to 10,000 g. Radiographs were taken of each subject's skull during exposure to load and used to determine the amount of separation at the coronal suture. Skull deflection resulting from most of the forces was not significant. However, the subject that experienced the additional loads between 100 g and 10,000 g showed separation of the coronal suture up to 0.31 mm at 5000 g and 0.91 mm at 10,000 g (Table 1).

Human studies

In 1971, Frymann[39] investigated cranial motion in living human subjects using a noninvasive apparatus that mechanically measured changes in cranial diameter. Frymann[39] did not report the number, age, or gender of the human subjects. In this study, subjects placed their heads in a U-shaped frame which had metallic rods attached to differential transformers. The metallic rods were then placed against the lateral aspects of the cranium. The differential transformers were used to measure the displacement of the metal rods, thereby measuring changes in the diameter of the cranium. Frymann[39] concluded that inherent cranial motion existed in the living humans that were tested, and that the cranial motion occurred in a rhythmic pattern that was slower than the cardiac and respiration rates. Although Frymann[39] provided example data plots of these measurements, the scale was not indicated. Therefore, quantitative measurements of cranial motion could not be determined in terms of micrometers (μm).

Pitlyk et al.[36] observed minute cranial motion of a dry human skull and a fresh human cadaver skull due to an externally applied load and motion of a fresh human cadaver skull due to increased intracranial pressure, with the skin and intracranial contents intact. Strain gauges were attached to an apparatus that was placed across the medial-lateral diameter of the skull to measure changes in diameter. Cranial deflections resulted from both externally applied force to the outside of the dry skull (direction of force not specified), and increased intracranial pressure in the cadaver. Intracranial pressure was increased by intraspinal injections of saline. It was determined that there is a correlation between the amount of saline injected into the spine and the amount of cranial motion. Unfortunately, the quantitative measurements of cranial motion were only reported in terms of the millivolt (mV) output of the strain gauges, as opposed to micrometers (μm) (Table 3). However, the data from this publication support the existence of cranial motion resulting from both externally applied force and increased intracranial pressure.

Table 3. Cranial movement in various human specimens.

Author	Subject Type	Number of Subjects	Spontaneous Motion vs. Applied Force	Observed Skull Deflection (Yes/ No)	Skull Deflection (μm)
Frymann[39]	Living Volunteers	N/R	Spontaneous Motion	Yes	N/D/R
Pitlyk et al.[36]	Dry Cadaver Skull	1	External Compressive Force	Yes	N/D/R
	Fresh Cadaver Skull	1	Increased Intracranial Pressure	Yes	N/D/R
Heifetz and Weiss[40]	Comatose Patient	2	Increased Intracranial Pressure	Yes	Subject 1 = 3.72 Subject 2 = 0.78

Note: N/R is Not Reported; N/D/R is No Deflection Values Reported.

Table options

Heifetz and Weiss[40] used an apparatus instrumented with strain gauges placed to measure changes across the medial-lateral diameter of the skull due to an increase in intracranial pressure in two comatose patients. The total intracranial pressure was elevated to between 15 mmHg and 20 mmHg using a different method in each patient. In Case 1, pressure was raised via bilateral jugular compression. In

Case 2, the cranial ventricles were infused with 7–12 cc of Ringer's lactate solution. The average skull deflections detected were 3.72 μm in Case 1 and 0.78 μm in Case 2 (Table 3).

Part II: review of human skull stiffness

Several studies have investigated the stiffness characteristics of the human skull by region and as a whole. The publications reviewed in this study utilized different methods to quantify human skull stiffness such as: mechanical impedance, impact tests, and drop tests.

Three studies[41, 42 and 43] determined skull stiffness through mechanical impedance techniques. A force generator was attached directly to a cadaver skull to exert low-amplitude sinusoidal forces of varying frequencies at the point of contact. The motion of the skull and forces exerted on the skull were recorded. The response of the skull varied with respect to frequency. Two frequencies of interest were those at anti-resonance, where skull stiffness is the greatest, and resonance, where the skull stiffness is the lowest. Although manual skull manipulation is performed at much lower frequencies than those corresponding to resonance and anti-resonance, the skull stiffness at anti-resonance is most applicable to the current study given that the stiffness at resonance is essentially zero.

Franke[41] performed mechanical impedance tests between 200 Hz and 1600 Hz on various human specimens. It was determined that a dry, eviscerated human skull had a higher resonant frequency than the same skull filled with gelatin. Franke[41] did not report the gender of the skull or if it was fresh versus embalmed. Tests were also performed on a human cadaver. Franke[41] did not report the gender of the cadaver or if it was fresh versus embalmed. The frequency corresponding to minimum stiffness (resonance) of the cadaver head was at 900 Hz. No maximum stiffness values (anti-resonance) or their corresponding frequencies were reported (Table 4). Finally, tests were performed on live subjects. Franke[41] did not report the age, gender, or number of the live subjects evaluated. No bone vibration was detected in the live human subjects, due to the poor coupling between the piston and skull. Minimum stiffness values for all subjects were very close to zero.

Table 4. Results of mechanical impedance studies on human specimens.

Author	Subject Type	Skull Region	Number of Specimens	Frequency at Maximum Stiffness [Anti-Resonance] (Hz)	Maximum Stiffness [Anti-Resonance] (N/mm)
Franke[41]	Dry Skull	Whole	1	N/R	N/R
	Dry Skull with Gelatin	Whole	1	N/R	N/R
	Cadaver	Whole	1	N/R	N/R
	Live humans	Whole	N/R	No bone vibration detected	
Hogdson et al.[42]	Cadaver	Frontal	2	360	26,269
		Side parietal	2	450	26,269
		Top parietal	2	300	15,761
		Occipital	2	180	29,772

Author	Subject Type	Skull Region	Number of Specimens	Frequency at Maximum Stiffness [Anti-Resonance] (Hz)	Maximum Stiffness [Anti-Resonance] (N/mm)
Stalnaker et al.[43]	Cadaver	Whole	1	166	4553

Note: N/R is Not Reported; N/D/R is No Deflection Values Reported.

Table options

A study conducted by Hodgson et al.[42] quantified the dynamic stiffness of the frontal, side parietal, top parietal, and occipital regions of two human skulls over a frequency range of 5 Hz–5000 Hz. Hodgson et al.[42]did not report the gender of the skull or if it was fresh versus embalmed. Maximum stiffness and the corresponding frequencies varied by region (Table 4). Frequencies corresponding to minimum stiffness also varied by region, but the corresponding skull stiffness values were essentially zero for all regions.

Stalnaker et al.[43] used mechanical impedance techniques to quantify skull stiffness of a fresh, unembalmed, male human cadaver head between frequencies of 30 Hz and 5000 Hz. Maximum stiffness was reported to occur at 166 Hz with a corresponding skull stiffness of 4553 N/mm (Table 4). Minimum stiffness was reported to occur at 820 Hz, but no corresponding minimum stiffness value was reported.

Other studies investigated skull stiffness by conducting compression tests,[44] impact tests,[45, 46, 47 and 49]and drop tests.[48] During all types of tests, the force exerted on the skull and the amount of skull deflection was measured. These data were used to generate applied force versus measured deflection curves (force–deflection curves) from which skull stiffness values were determined. Force–deflection curves often exhibit a nonlinear toe region in which the skin deflects first and then the skull (Fig. 1). The majority of the studies reviewed in the current study did not include the toe region for skull stiffness calculations, but used only the linear portion of the curve.[44, 45, 46, 47 and 49] There was only one study which included the nonlinear toe region by using the peak force and peak deflection to calculate skull stiffness.[48]

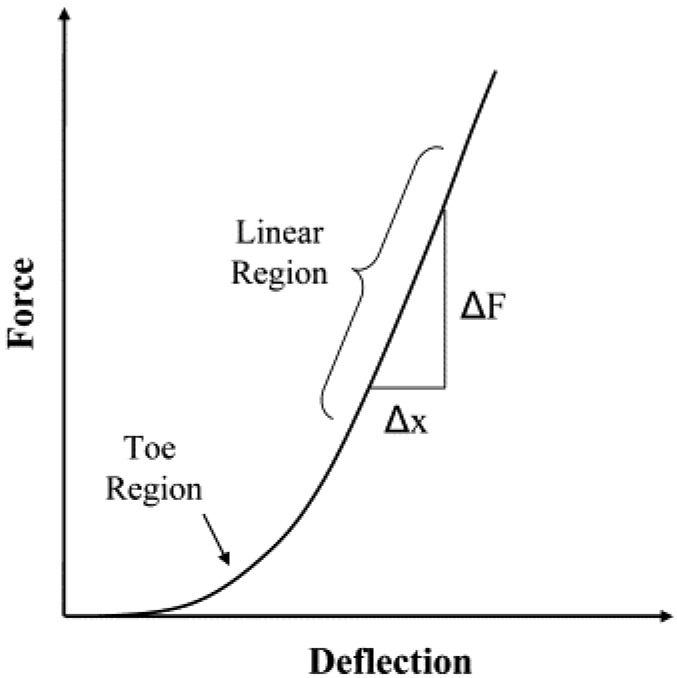


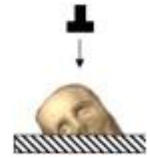
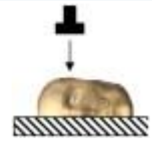
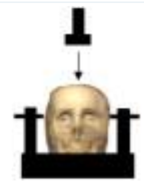
Figure 1. Typical force–deflection curve.

Figure options

McElhaney et al.[44] measured the force-deflection response of twenty-four unembalmed human cadavers heads (20 male, 3 female) positioned between two steel platens (150 mm diameter). A total of twelve tests were performed by loading the frontal and occipital bone (anterior-posterior compression), and twelve tests were performed by loading the right and left temporo-parietal regions (right-left compression). Stiffness values were calculated from the linear-most region of the force–deflection curve as demonstrated in Fig. 1. McElhaney et al.[44] reported that the stiffness for the frontal region ranged from 1400 to 3500 N/mm while the stiffness in the temporal region ranged from 700 to 1750 N/mm (Table 5).

Table 5. Human skull stiffness values determined by compression tests, impact tests, and drop tests.

Author	Skull Loading Bone/ Region	Loading Rate (m/s)	Number of Specimens	Average Stiffness [Min-Max] (N/mm)	Test Setup
McElhaney et al.[44]	Frontal- Occipital (Anterior-Posterior) Circular Disks	N/R	12	1225 [700–1750]	
	Temporo-Parietal (Right-Left) Circular Disks	N/R	12	2450 [1400–3500]	
Allsop et al.[45]	Frontal Rectangular Impactor	N/R	13	1013 [400–2220]	

Author	Skull Loading Bone/ Region	Loading Rate (m/s)	Number of Specimens	Average Stiffness [Min-Max] (N/mm)	Test Setup
Allsop et al.[46]	Parietal Rectangular Impactor	4.3	11	4200 [1600–6430]	
	Temporo-Parietal Circular Impactor	2.7	20	1800 [700–4760]	
Yoganandan et al.[47]	Average of top, parietal, temporal, frontal, occipital Circular Impactor	0.0025	6	812 [467–1290]	
	Average of top, parietal, temporal, frontal, occipital Circular Impactor	7.1–8.0	6	4023 [2462–5867]	
Yoganandan et al.[48]	Temporo-Parietal Drop Test	4.9–7.7	10	562 [390–689]	
Cormier et al.[49 and 94]	Frontal Circular Impactor	5.3	26	475 [not specified]	

Note: N/R is Not Reported.

Table options

A study conducted by Allsop et al.[45] investigated frontal bone impact and stiffness. Thirteen unembalmed male cadaver heads (4 male, 9 female), including the skin and intracranial contents, were exposed to impacts with a semi-circular, 230 mm long impactor. Force exerted on the skull was measured with a series of miniature load cells. A potentiometer was used to measure the deflection of the skull. Stiffness values were calculated from the linear-most region of the force–deflection curve. The average stiffness of the frontal bone was reported to be 1013 N/mm (Table 5).

A second study conducted by Allsop et al.[46] measured force-deflection characteristics of the temporo-parietal region of the skull on thirty-one (12 male, 19 female) unembalmed human cadaver heads, including the skin and intracranial contents. Cadaver heads were mounted in plaster and impacted by one of two impactors: a flat, rectangular surface; or a flat, circular impacting surface (Table 5). The impactors were instrumented with force transducers and string potentiometers to measure force on the skull and deflection of the skull, respectively. The rectangular impactor was dropped onto the parietal region of the skull from a height of 102 cm, which resulted in an impact velocity of 4.3 m/s. The circular impactor was dropped onto the temporo-parietal region from a height of 38 cm and impacted the head at a velocity of 2.7 m/s. Stiffness values were calculated from the linear-most region of the force–deflection curve. The average stiffness was 4200 N/mm for impacts with the rectangular impactor and 1800 N/mm for impacts with the circular impactor (Table 5).

Yoganandan et al.[47] determined the stiffness characteristics of the entire skull at two loading rates. Twelve unembalmed human cadaver heads (5 male, 7 female), with the skin and intracranial contents intact, were exposed to a low speed loading rate of 0.0025 m/s and high speed loading rates between 7.1 m/s and 8.0 m/s. All impacts were performed with a hemispherical anvil (Table 5). Several regions of the skull were tested. The force exerted on the skull was measured by a uniaxial force gauges while the deflection of the skull was measured by a linear variable differential transformer. Stiffness values were determined from the linear-most region of force–deflection curves. For the low speed tests, stiffness ranged from 467 N/mm to 1290 N/mm with an average of 812 N/mm (Table 5). For the high speed tests, stiffness ranged from 2462 N/mm to 5867 N/mm with an average of 4023 N/mm (Table 5).

A second study conducted by Yoganandan et al.[48] investigated the effects of lateral impacts on the skull. Ten unembalmed human cadaver heads, with the skin and intracranial contents intact, were tested. Yoganandan et al.[48] did not report the gender of the cadavers. The skulls were instrumented with tri-axial accelerometers to calculate skull deflection at the temporo-parietal region contralateral to the impacted side, the anterior region, and the posterior region of the head. The heads were dropped from various heights, with impact velocities between 4.9 m/s and 7.7 m/s. The skulls were oriented so that the temporo-parietal region of the skull impacted a padded force platform which was instrumented with a six-axis load cell to measure the force exerted on the skull (Table 5). The average skull stiffness was 562 N/mm (Table 5). Using the data presented in this study, skull stiffness was calculated using the peak force and peak deflection and therefore included the toe region of the force–deflection curve. Consequently, the reported skull stiffness may be lower than other studies which do not include the toe region in the stiffness calculation.

A study conducted by Cormier et al.[49 and 94] investigated frontal bone impact and stiffness. Twenty-seven unembalmed male cadaver heads, including the skin and intracranial contents, were exposed to forty-six impacts with a rigid cylindrical impactor. The impact speed was 5.3 m/s. A load cell was used to measure the force exerted on the skull. Two uniaxial accelerometers were used to calculate the deflection of the skull. Stiffness values were calculated from the linear-most region of the force–deflection curve. The average stiffness of the frontal bone was reported to be 475 N/mm (Table 5).

Part III: biomechanical analysis

The amount of external force required to achieve the human skull deflections reported in the literature was estimated using the average human skull stiffness values reported in Table 5. This was done by multiplying the average skull stiffness (N/mm) for a given study by the skull deflections (mm) reported in the literature (Table 6). Since Heifetz and Weiss[40] were the only researchers to report cranial deflections of human specimens, cranial deflections from this study were used estimate the amount of external force required to achieve skull deflections in living humans. For adult human skulls, the forces ranged from 0.44 N to 23.22 N for cranial movement of 0.78 μm and from 1.78 N to 110.75 N for cranial movement of 3.72 μm.

Table 6. Estimated force required to achieve the cranial deflections reported by Heifetz and Weiss.[40]

Study	Skull Region/Bone	Test Method	Case 1: 3.72 μm	Case 2: 0.78 μm
			Required Force (N)	
Hogdson et al.[42]	Frontal Bone	Mech. Impedance	97.72	20.49
	Side Parietal Bone	Mech. Impedance	97.72	20.49
	Top Parietal Bone	Mech. Impedance	58.63	12.29
	Occipital Bone	Mech. Impedance	110.75	23.22
Stalnaker et al.[43]	Whole Skull	Mech. Impedance	16.94	3.55
McElhaney et al.[44]	Frontal-Occipital	A-P Compression	4.56	0.96
	Temporo-Parietal	R-L Compression	9.11	1.91
Allsop et al.[45]	Frontal Bone	Rectangular Impactor	3.77	0.79
Allsop et al.[46]	Parietal Bone	Rectangular Impactor	15.62	3.28
	Temporo-Parietal	Circular Impactor	6.70	1.40
Yoganandan et al.[47 and 48]	Average – All Regions (Low Speed)	Circular Impactor	3.02	0.63
	Average – All Regions (High Speed)	Circular Impactor	21.83	4.58
	Temporo-Parietal	Drop Test	2.09	0.44
Cormier et al.[49 and 79]	Frontal Bone	Circular Impactor	1.78	0.37
		Maximum	**110.75**	**23.22**
		Minimum	**1.78**	**0.44**
		Average	**32.16**	**6.74**

Table options

Discussion

The purpose of this study was to review findings of cranial movement in animal and human specimens and evaluate the validity of cranial movement due to manual manipulation in humans through biomechanical analysis. Several studies reported quantitative and qualitative results of cranial movement. In these studies, cranial motion was induced by various internal and external stresses on the cranium. A wide range of cranial motion was observed across animal and human specimens. A biomechanical analysis was conducted to determine whether measurements of human cranial motion as a result of CO were reasonable.

The review of literature regarding cranial motion in animals illustrated that measurable cranial motion across the sagittal and coronal sutures does occur in both adolescent and adult mammalian species due to externally applied forces and increases in intracranial pressure.[33, 34, 35, 36, 37 and 38] The finding that motion across cranial sutures is possible is further supported by studies which have investigated the properties of isolated bone-suture-bone specimens. These studies have reported that the cranial sutures are not only compliant, but are significantly weaker and more compliant than the surrounding cranial bone.[50, 51, 52,53 and 54] This finding implies that deflection across the suture will occur before deflection of the cranial bone. Although the strength of the cranial sutures has been reported to be positively correlated to the degree of bone interdigitation, the strength of the sutures never exceeded that of the cranial bone.[50] In regard to adolescent skulls versus adult skulls, studies on isolated bone-suture-bone specimens have shown that adult human cranial sutures are stiffer[51] than the cranial sutures of human infants.[53 and 54] In addition, adult human cortical bone is stiffer than the cortical bone in human children.[53, 54, 55, 56, 57, 58, 59 and 60] Therefore, the overall stiffness of the human skull increases with age, which could potentially explain the success of CO in treating the symptoms of neurological and medical problems in children.[1, 2 and 3]

The review of literature regarding cranial motion in humans illustrated that measurable changes in cranial vault diameter (medial-lateral) does occur in adult humans due to both externally applied forces and increases in intracranial pressure.[36, 39 and 40] This was found to be true in both post-mortem human skulls[36]and living human subjects.[39 and 40] The finding that cranial deflection does occur in human skulls is further supported by recent publications which, through the use of medical imaging modalities, reported deformations of the falx cerebri following externally applied forces to the skull[61] and changes in the geometry of the cranial vault following CO.[62]

The review of literature regarding the stiffness of the human skull showed that skull stiffness values varied considerably between studies (390 N/mm to 6430 N/mm). Given the complex geometry of the human skull and cranial sutures, some of the variation could be attributed to differences in skull stiffness by region (i.e. frontal, occipital, parietal, temporo-parietal, etc.) and direction of load application. Some of the variation could also be attributed to variation between subjects. It has been well established in the literature that the structural response of bone varies considerably between individuals due to differences in age, gender, bone mineral density, material properties, geometry, and cortical bone thickness.[59, 63, 64, 65, 66, 67, 68, 69, 70, 71,72 and 73] Finally, differences in loading area and loading velocity can result in differences in measured response. For example, a larger force would be required to obtain a given amount of deflection when using a large loading area versus a small loading area (e.g.: a large plate versus a needle). Therefore, measured stiffness can vary with respect to the area of loading in a given experiment. With respect to loading velocity, it is well known that bone is a viscoelastic material and is therefore affected by the rate of loading.[56, 74, 75,76 and 77] Assuming that all other test conditions are similar, the stiffness of bone will increase with increased loading velocity. Given the number of potential reasons for the variation in reported skull stiffness between studies, it is not possible to determine the relative contributions of each of variable without conducting a controlled study. Regardless, these studies provide valuable data regarding the range of human skull stiffness values under various loading conditions.

To estimate the amount of external force required to induce reported cranial deflections in human specimens, the measured human skull deflections from published studies[36, 39 and 40] were used in conjunction with published human skull stiffness values.[41, 42, 43, 44, 45, 46, 47, 48 and 49] Since Pitlyk et al.[36] and Frymann[39] did not report the quantitative measurements of cranial deflection in micrometers (μm), no forces could be calculated from these studies. Heifetz and Weiss[40] reported small cranial deflections (0.78 μm and 3.72 μm) due to increases in intracranial pressure in two comatose patients. Depending on the region of the head to which forces are applied, it was estimated that a force ranging from 0.44 N to 23.22 N would be required to cause 0.78 μm of cranial deflection and 2.09 N–111.75 N for cranial deflections of 3.72 μm. For comparison, CO palpation studies have reported that the force applied by the index finger during CO ranges between 0.13 N and 0.80 N during frontal bone palpation and between 0.8 N and 8.2 N during cervical spine palpation.[78 and 79] Given that osteopathic physicians typically use more than one finger during manual manipulation, the total amount of force applied to the skull during CO would be greater (up to 5 times greater or 0.65 N–41 N) than the forces reported in the CO palpation studies. Therefore, the total force likely exerted on the skull during CO would lie within the range of forces required to cause the cranial deflections reported in the literature. The finding that cranial motion due to CO can occur in human skulls is further supported by a recent publication which, through the use of medical imaging modalities, reported changes in the geometry of the cranial vault following CO.[52]

Limitations and future studies

There are limitations in the current study that should be addressed. Many of the cranial motion investigations reviewed in the current study involve expansion of the skull while CO involves compression of the skull. The accuracy of the reported measurements of cranial deflection must also be considered. The forces quantified in the current study are based on only two subjects in whom very small amounts of cranial deflection were measured.[21] Such measurements would require extremely sensitive strain gauges and an extremely stiff test apparatus. Since little is known about the strain gauges used in the study, the accuracy and sensitivity of those strain gauges could not be assessed. In addition, the strain gauges used to measure cranial deflection on those subjects measured changes across the diameter of the head. The expansion or contraction of the cranial vault could be a result of motion at the sutures, flexing of the cranial bones, or a combination of both.[23] Therefore, the cranial deflection measurements reported by Heifetz and Weiss[40]cannot be attributed solely to motion at the sutures, if at all.

Future studies should be conducted to determine the relative amount of motion at the sutures versus cranial bone flexion during expansion or contraction of the cranial vault. This could be done experimentally with the apparatus used by Heifetz and Weiss[40] in conjunction with strain gauges placed across the relevant cranial sutures. This limitation could also be addressed with the use of computational modeling. Finite element models of the human body have advanced dramatically over recent years, and are extensively validated using relevant biomechanical data. These models are commonly used to estimate hard and soft tissue deformations and assess injury risk due to externally applied loading.[80, 81, 82, 83, 84, 85, 86, 87, 88, 89, 90, 91, 92 and 93] Therefore, existing human head models[89, 90, 91, 92 and 93] could be used to quantify the amount of motion at the sutures versus cranial bone flexion resulting from external forces applied during CO with the use of published cranial bone material properties,[53, 54, 55, 56, 57 and 58] isolated suture properties,[50, 51, 52, 53 and 54] and whole skull response data.[41, 42, 43, 44, 45, 46, 47, 48 and 49]

The relevance of this study to CO should also be considered. Although CO has been used to treat symptoms of medical and neurological problems, the underlying physiological mechanisms of CO must be investigated. In other words, the correlation between cranial motion and the effectiveness of CO should be investigated. Although the results of the current study indicate that cranial deflection can be generated due to the forces applied during CO, there is no existing evidence as to whether cranial motion contributes to the reduction in symptoms that result from CO. Patient improvement may be attributed to several different mechanisms such as: a placebo effect, relaxation of muscles due to massaging effects, or a decrease in pressure on the cranial nerves due to massaging. Therefore, further research should be conducted to investigate the effects of cranial motion in CO on patient improvement.

Conclusions

In the current study, the validity of cranial movement due to manual manipulation was evaluated through a review of the literature and a biomechanical analysis to determine the amount of force necessary to generate reported cranial deflections in humans. The review of literature regarding cranial motion illustrated that both externally applied forces and increases in intracranial pressure result in measurable motion across the cranial sutures in adolescent and adult mammalian species, and measurable changes in cranial vault diameter in post-mortem and living adult human skulls. However, the amount of cranial motion may vary by subject, the region of the head to which forces are applied, and the method of force application. For living humans, cranial vault deflections were reported to range from 0.78 μm to 3.72 μm. However, it is unclear as to whether the expansion of the cranial vault was a result of motion at the sutures, flexing of the cranial bones, or a combination of both. The review of literature regarding the stiffness of the human skull showed that skull stiffness values ranged from 390 N/mm to 6430 N/mm depending on the region of the skull and method of loading. Based on the range of skull stiffness values, it was determined that an applied force between 0.44 N and 23.2 N would be required to cause 0.78 μm of deflection, and between 2.09 N and 111 N would be required to cause 3.72 μm of deflection. Given that the forces required to generate reported cranial deflections in living humans are within the range of forces likely to be used during CO, it is reasonable that small amounts of cranial deflection can occur as a result of the forces applied to the skull during CO.

Author contribution

All of the authors listed on the current manuscript were involved in the study, preparation of the manuscript, and approval of the final version submitted.

Conflict of interest statement

There are no financial or personal relationships with other people or organizations that could inappropriately influence the work presented in the current study.

References

1.
    - 1
    - V.M. Frymann, R.E. Carney, P. Springall
    - **Effect of osteopathic medicine on neurologic development in children**
    - J Am Osteopathic Assoc, 92 (1992), pp. 729–744
    - View Record in Scopus
      | Cited By in Scopus (14)
2.
    - 2
    - C. Hayden, B. Mullinger
    - **A preliminary assessment of the impact of cranial osteopathy for the relief of infantile colic**
    - Complement Therapies Clin Practices, 12 (2006), pp. 83–90
    - **Article**
      |
      PDF (156 K)
      |
      View Record in Scopus
      | Cited By in Scopus (26)
3.
    - 3
    - P. Joyce, C. Clark
    - **The use of craniosacral therapy to treat gastroesophageal reflux in infants**
    - Infants Young Child, 9 (1996), pp. 51–57
    -
4.
    - 4

- M.E. Sandhouse, D. Shechtman, R. Sorkin, J.L. Drowos, A.J. Caban-Martinez, M.M. Patterson *et al.*
- **Effect of osteopathy in the cranial field on visual function – a pilot study**
- J Am Osteopathic Assoc, 110 (2010), pp. 239–243
- View Record in Scopus
  | Cited By in Scopus (4)

5.
- 5
- L. Barke, S. Gelman, J.A. Lipton
- **A successful use of cranial-sacral osteopathy in the treatment of post-traumatic headache following subarachnoid hemorrhage**
- Am Acad Osteopathy J (1997), pp. 22–23
-

6.
- 6
- K. Shain, M.L. Madigan, S. Rowson, J. Bisplinghoff, S.M. Duma
- **Analysis of the ability of catcher's masks to attenuate head accelerations on impact with a baseball**
- Clin J Sport Med, 20 (2010), pp. 422–427
- View Record in Scopus
  |
  Full Text via CrossRef
  | Cited By in Scopus (10)

7.
- 7
- S.M. Duma, S. Rowson
- **Past, present, and future of head injury research**
- Exerc Sport Sci Rev, 39 (2011), pp. 2–3
- View Record in Scopus
  |
  Full Text via CrossRef
  | Cited By in Scopus (14)

8.
- 8
- S. Rowson, S.M. Duma

- **Development of the star evaluation system for football helmets: integrating player head impact exposure and risk of concussion**
- Ann Biomed Eng, 39 (2011), pp. 2130–2140
- View Record in Scopus
  |
  Full Text via CrossRef
  | Cited By in Scopus (25)

9.
- 9
- S. Rowson, P.G. Brolinson, M.W. Goforth, D. Dietter, S.M. Duma
- **Linear and angular head accelerations measurements in collegiate football**
- J Biomechanical Eng, 131 (2009), p. 061016
- View Record in Scopus
  |
  Full Text via CrossRef
  | Cited By in Scopus (34)

10.
- 10
- J.J. Crisco, B.J. Wilcox, J.G. Beckwith, J.J. Chu, A.C. Duhaime, S. Rowson *et al.*
- **Head impact exposure in collegiate football players**
- J Biomech, 44 (2011), pp. 2673–2678
- Article
  |
  PDF (309 K)
  |
  View Record in Scopus
  | Cited By in Scopus (7)

11.
- 11
- S. Rowson, S.M. Duma, J.G. Beckwith, J.J. Chu, R.M. Greenwald, J.J. Crisco *et al.*
- **Rotational head kinematics in football impacts: an injury risk function for concussion**
- Ann Biomed Eng, 40 (2012), pp. 1–13
- View Record in Scopus
  |

Full Text via CrossRef
| Cited By in Scopus (13)

12.
- 12
- J.A. Beyer, S. Rowson, S.M. Duma
- **Concussions experienced by major league baseball catchers and umpires: field data and experimental baseball impacts**
- Ann Biomed Eng, 40 (2012), pp. 150–159
- View Record in Scopus
  |
  Full Text via CrossRef
  | Cited By in Scopus (4)

13.
- 13
- E.G. Takhounts, S.A. Ridella, V. Hasija, R.E. Tannous, J.Q. Campbell, D. Malone *et al.*
- **Investigation of traumatic brain injury using the next generations of simulated injury monitor (SIMon) finite element head model**
- Stapp Car Crash J, 52 (2008), pp. 1–31
- View Record in Scopus
  | Cited By in Scopus (35)

14.
- 14
- W. Rutland-Brown, J.A. Langlois, K.E. Thomas, Y.L. Xi
- **Incidence of traumatic brain injury in the United States, 2003**
- J Head Trauma Rehabil, 21 (2006), pp. 544–548
- View Record in Scopus
  |
  Full Text via CrossRef
  | Cited By in Scopus (194)

15.
- 15
- S.M. Duma, J.R. Crandall, R.W. Rudd, R.W. Kent
- **Small female head and neck interaction with a deploying side airbag**
- Accidental Anal Prev, 35 (2003), pp. 811–816

- Article
  |
  PDF (259 K)
  |
  View Record in Scopus
  | Cited By in Scopus (21)

16.
- 16
- A.M. Eigen, P.G. Martin
- **Identification of real world injury patterns in aid of dummy development**
- 19th Enhanced Safety of Vehicles (2005)
-

17.
- 17
- J.H. Stuhmiller, Y.Y. Phillips III, D.R. Richmond
- **Conventional warfare: ballistic, blast, and burn, injuries. [Chapter 7]-The physics and mechanisms of primary blast injury**
- Bordeninstitute.army.mil (1991)
-

18.
- 18
- M.A. Mayorga
- **The pathology of primary blast overpressure injury**
- Toxicology (1997), pp. 17–28
- Article
  |
  PDF (7707 K)
  |
  View Record in Scopus
  | Cited By in Scopus (110)

19.
- 19
- A.D. Leonardi, C.A. Bir, D.V. Ritzel, P.J. VandeVord
- **Intracranial pressure increases during exposure to a shock wave**

- J Neurotrauma, 28 (2011), pp. 85–94
- View Record in Scopus
  |
  Full Text via CrossRef
  | Cited By in Scopus (16)

20.
- 20
- W.G. Sutherland
- **The cranial bowl**
- Free Press, Mankato, MN (1939)
-

1.
- 21
- H.H. King
- **Osteopathy in the cranial field**
- Anthony G. Chila (Ed.), Foundations of osteopathic medicine (3rd ed.), Lippincott Williams & Wilkinson, Baltimore MD (2011), pp. 728–748
-

2.
- 22
- H.I. Magoun
- **Osteopathy in the cranial field, 2e**
- Journal Publishing Company, Kirksville, MO (1966) 112–113
-

3.
- 23
- J.S. Rogers, P.L. Witt
- **The controversy of cranial bone motion**
- JOSPT, 26 (1997), pp. 95–103
-

4.
- 24
- A. Ferguson
- **A review of the physiology of cranial osteopathy**

- J Osteopathic Med, 6 (2003), pp. 74–88
-

5.

- 25
- F.H. Netter
- **Atlas of human anatomy**
- (4th ed.)Saunders-Elsevier, Philadelphia, PA (2006)
-

6.

- 26
- L.A. Madeline, A.D. Elster
- **Suture closure in the human chondrocranium: CT assessment**
- Radiology, 196 (1995), pp. 747–756
-

7.

- 27
- K. Okamoto, J. Ito, S. Tokiguchi, T. Furusawa
- **High-resolution CT findings in the development of the sphenooccipital synchondrosis**
- Am J Neuroradiol, 17 (1996), pp. 117–120
-

8.

- 28
- L. Flores-Sarnat
- **New insights into craniosynostosis**
- Semin Pediatr Neurol, 9 (2002), pp. 274–291
- Article
  |
  PDF (6758 K)
  |
  View Record in Scopus
  | Cited By in Scopus (16)

9.

- 29
- H.B. Sarnat, J.H. Menkes

- **Neuroembriology, genetic programming, and malformations of the nervous system**
- J.H. Menkes, H.B. Sarnat (Eds.), Child neurology (6th ed.), Lippincott Williams & Wilkins (2000), pp. 351–354
-

10.
- 30
- C. Green, C. Martin, K. Bassett, A. Kazanjian
- **A systematic review and critical appraisal of the scientific evidence on craniosacral therapy**
- British Columbia Office of Health Technology Assessment, The University of British Columbia (1999)
-

11.
- 31
- V.G. Kokich
- **Age changes in the human frontozygomatic suture from 20 to 95 years**
- Am J Orthodont, 69 (1976), pp. 411–430
- **Article**
  |
  PDF (12160 K)
  |
  View Record in Scopus
  | Cited By in Scopus (33)

12.
- 32
- E.W. Retzlaff, J. Upledger, F.L. Mitchell, J. Walsh
- **Aging of cranial sutures in humans**
- Anat Rec, 193 (1979), p. 663 [abstract]
-

13.
- 33
- D.K. Michael, E.W. Retzlaff
- **A preliminary study of cranial bone movement in squirrel monkey**
- J Am Osteopath Assoc, 74 (1975), pp. 866–869
- View Record in Scopus
  | Cited By in Scopus (9)

14.

- 34
- E.W. Retzlaff, D.K. Michael, R. Roppel
- **Cranial bone mobility**
- J Am Osteopath Assoc, 74 (1975), pp. 869–873
- View Record in Scopus
 | Cited By in Scopus (9)

15.

- 35
- H.A. Oudhof, W.J. van Doorenmaalen
- **Skull morphogenesis and growth: hemodynamic influence**
- Acta Anatomica (1983), pp. 181–186
- View Record in Scopus
 |
 Full Text via CrossRef
 | Cited By in Scopus (10)

16.

- 36
- P.J. Pitlyk, T.P. Piantanida, D.W. Ploeger
- **Noninvasive intracranial pressure monitoring**
- Neurosurgery (1985), pp. 581–584
- View Record in Scopus
 |
 Full Text via CrossRef
 | Cited By in Scopus (6)

17.

- 37
- T. Adams, R.S. Heisey, M.C. Smith, B.J. Briner
- **Parietal bone mobility in the anesthetized cat**
- J Am Osteopathic Assoc (1992), pp. 599–622
-

18.

- 38
- P.A. Downey, T. Barbano, R. Kapur-Wadhwa, J.J. Sciote, M.I. Siegel, M.P. Mooney

- **Craniosacral therapy: the effects of cranial manipulation on intracranial pressure and cranial bone movement**
- J Orthopaedics Sports Phys Ther (2006), pp. 845–853
- View Record in Scopus
  |
  Full Text via CrossRef
  | Cited By in Scopus (10)

19.
- 39
- V.M. Frymann
- **A study of rhythmic motions of the living cranium**
- J Am Ostepath Assoc, 70 (1971), pp. 928–945
- View Record in Scopus
  | Cited By in Scopus (32)

20.
- 40
- M.D. Heifetz, M. Weiss
- **Detection of skull expansion with increased intracranial pressure**
- J Neurosurg (1981), pp. 811–812
- View Record in Scopus
  |
  Full Text via CrossRef
  | Cited By in Scopus (14)

1.
- 41
- E.K. Franke
- **Response of the human skull to mechanical vibrations**
- J Acoust Soc America, 28 (1956), pp. 1277–1284
- View Record in Scopus
  |
  Full Text via CrossRef
  | Cited By in Scopus (16)

2.
- 42

- V.R. Hodgson, E.S. Gurdjian, L.S. Thomas
- **The determination of response characteristics of the head with emphasis on mechanical impedance techniques. Eleventh Stapp Car Crash Conference**
- Society of Automotive Engineers, Inc., New York (1967) 125–138
-

3.
- 43
- R.L. Stalnaker, J.L. Fogle, J.H. McElhaney
- **Driving point impedance characterisitcs of the head**
- J Biomech, 4 (1971), pp. 127–139
- Article
  |
  PDF (2467 K)

4.
- 44
- J.H. McElhaney, R.L. Stalnaker, V.L. Roberts
- **Biomechanical aspects of head injury**
- W. King, H. Mertz (Eds.), Human impact tolerance, Plenum Press, New York (1972), pp. 85–112
-

5.
- 45
- D.L. Allsop, C.Y. Warner, M.G. Wille, D.C. Schneider, A.M. Nahum
- **Facial impact response – a comparison of the hybrid III dummy and human cadaver**
- 32nd Stapp Car Crash Conference (1988), pp. 17–19 SAE Paper No. 881719
- View Record in Scopus
  | Cited By in Scopus (2)

6.
- 46
- D.L. Allsop, T.R. Perl, C.Y. Warner
- **Force/Deflection and fracture characteristics of the temporo-parietal region of the human head**
- 35th Stapp Car Crash Conference, Society of Automotive Engineers, San Diego, California (1991)
-

7.

- 47
- N. Yoganandan, F.A. Pintar, A. Sances, P.R. Walsh, C.L. Ewing, D.J. Thomas *et al.*
- **Biomechanics of skull fracture**
- J Neurotrauma (1995), pp. 659–668
- View Record in Scopus
  |
  Full Text via CrossRef
  | Cited By in Scopus (53)

8.
- 48
- N. Yoganandan, J. Zhang, S. Kuppa, R.H. Eppinger
- **Biomechanics of lateral skull fracture**
- International Research Council on the Biomechanics of Impact, Lisbon, Portugal (2003) 69–78
-

9.
- 49
- J.M. Cormier, S.J. Manoogian, J.A. Bisplinghoff, S. Rowson, A.C. Santago, C. McNally *et al.*
- **The tolerance of the frontal bone to blunt impact**
- J Biomechanical Eng, 133 (2011), p. 021004
- View Record in Scopus
  |
  Full Text via CrossRef
  | Cited By in Scopus (1)

10.
- 50
- C.R. Jaslow
- **Mechanical properties of cranial sutures**
- J Biomech, 23 (1990), pp. 313–321
- **Article**
  |
  PDF (2830 K)
  |
  View Record in Scopus
  | Cited By in Scopus (89)

11.

- 51
- R.P. Hubbard
- **Flexure of layered cranial bone**
- J Biomechanics, 4 (1971), pp. 251–263
- Article
  |
  PDF (1078 K)
  |
  View Record in Scopus
  | Cited By in Scopus (30)

12.

- 52
- R.P. Hubbard, J.W. Melvin, I.T. Barodawala
- **Flexure of cranial sutures**
- J Biomechanics, 4 (1971), pp. 491–496
-

13.

- 53
- B. Coats, S.S. Margulies
- **Material properties of human infant skull and suture at high rates**
- J Neurotrauma, 23 (2006), pp. 1222–1232
-

14.

- 54
- S.S. Margulies, K.L. Thibault
- **Infant skull and suture properties: measurements and implications for mechanisms of pediatric brain injury**
- J Biomechanic Eng, 122 (2000), pp. 364–371
-

15.

- 55
- J.H. McElhaney, J.L. Fogle, J.W. Melvin, R.R. Haynes, V.L. Roberts, N.M. Alem
- **Mechanical properties on cranial bone**

- J Biomech, 3 (1970), pp. 495–511

16.

- 56
- J.L. Wood
- **Dynamic response of human cranial bone**
- J Biomechanics, 4 (1971), pp. 1–12

17.

- 57
- J.A. Motherway, P. Verschueren, G. Van der Perre, J. Vander Sloten, M.D. Gilchrist
- **The mechanical properties of cranial bone: the effect of loading rate and cranial sampling position**
- J Biomech, 42 (2009 Sep 18), pp. 2129–2135

18.

- 58
- G. McPherson, T. Kriewall
- **The elastic modulus of fetal cranial bone: a first step toward understanding of the biomechanics of fetal head molding**
- J Biomech, 13 (1980), pp. 9–16

19.

- 59
- A.R. Kemper, C. McNally, E.A. Kennedy, A. Rath, S.J. Manoogian, J.D. Stitzel *et al.*
- **Material properties of human rib cortical bone from dynamic tension coupon testing**
- Stapp Car Crash J, 49 (2005), pp. 199–230

20.

- 60
- A.H. Burstein, D.T. Reilly, M. Martens
- **Aging of bone tissue: mechanical properties**
- J Bone Jt Surg, 58-A (1976), pp. 82–86

1.

- 61
- D.C. Kostopoulos, G. Keramidas
- **Changes in elongation of falx cerebri during craniosacral therapy techniques applied on the skull of an embalmed cadaver**
- Cranio, 10 (1992), pp. 9–12
- View Record in Scopus
  | Cited By in Scopus (8)

2.
- 62
- S.L. Oleski, G.H. Smith, W.T. Crow
- **Radiographic evidence of cranial bone mobility**
- Cranio, 20 (2002), pp. 34–38
- View Record in Scopus
  | Cited By in Scopus (10)

3.
- 63
- R. Eppinger, J. Marcus, R. Morgan
- **Development of dummy and injury index for NHTSA's thoracic side impact protection research program**
- Society of Automotive Engineers, Warrendale, PA (1984) SAE paper number 840885
-

4.
- 64
- D. Kallieris, R. Mattern, A. McIntosh, F. Boggasch
- **The biofidelity of EuroSID 1 and BioSID**
- Proc. 36th Stapp Car Crash Conference, 36Society of Automotive Engineers, Warrendale, PA (1992), pp. 103–117
- View Record in Scopus
  | Cited By in Scopus (1)

5.
- 65
- J. Cavanaugh, Y. Huang, Y. Zhu, A. King
- **Injury response of the thorax in side impact cadaver tests**

- 37th Proc. Stapp Car Crash Conference, 37Society of Automotive Engineers, Warrendale, PA (1993), pp. 973–980
-

6.
   - 66
   - Q. Zhou, S. Rouhana, J. Melvin
   - **Age effects on thoracic injury tolerance**
   - Society of Automotive Engineers, Warrendale, PA (1996) SAE paper number 962421
   -

7.
   - 67
   - A.R. Kemper, C. McNally, C.A. Pullins, L.J. Freeman, S.M. Duma
   - **The biomechanics of human ribs: material and structural properties from dynamic tension and bending tests**
   - Stapp Car Crash J, 51 (2007), pp. 235–273
   - View Record in Scopus
     | Cited By in Scopus (24)

8.
   - 68
   - T.S. Keller
   - **Predicting to compressive mechanical behavior of bone**
   - J Biomech, 27 (1994), pp. 1159–1168
   -

9.
   - 69
   - D.R. Carter, W.C. Haynes
   - **The compressive behavior of bone as a two-phase porous structure**
   - J Bone Jt Surg, 59-A (1976), pp. 954–962
   -

10.
   - 70
   - R.B. Martin, J. Ishida
   - **The relative effects of collagen fiber orientation, porosity, density, and mineralization, on bone strength**

- J Biomech, 22 (1989), pp. 419–426
- Article
  |
  PDF (2447 K)
  |
  View Record in Scopus
  | Cited By in Scopus (128)

11.
- 71
- R.W. McCalden, J.A. McGeogh, M.B. Barker, C.M. Court-Brown
- **Age-related changes in the tensile properties of cortical bone**
- J Bone Jt Surg, 75 (1993), pp. 1193–1205
- View Record in Scopus
  | Cited By in Scopus (260)

12.
- 72
- K.H. Muellar, A. Trias, R.D. Ray
- **Bone density and composition. Age-related and phathological changes in water and mineralization content**
- J Bone Jt Surg, 48-A (1996), pp. 140–148
-

13.
- 73
- M.B. Schaffler, D.B. Burr
- **Stiffness of compact bone: effects of porosity and density**
- J Biomech, 21 (1988), pp. 13–16
- Article
  |
  PDF (466 K)
  |
  View Record in Scopus
  | Cited By in Scopus (191)

14.
- 74

- J. McElhaney, E. Byars
- **Dynamic response of biological materials**
- ASME (1965) Publl. 65-WA/Huf-9
-

15.
- 75
- R. Crowninshield, M. Pope
- **The response of compact bone in tension at various strain rates**
- Ann Biomed Eng, 2 (1974), pp. 217–225
- View Record in Scopus
  |
  Full Text via CrossRef
  | Cited By in Scopus (35)

16.
- 76
- T. Wright, W. Hayes
- **Tensile testing of bone over a wide range of strain rates: effects of strain rate, microstructure and density**
- Med Biol Eng (1976 Nov), pp. 671–679
-

17.
- 77
- D. Carter, W. Haynes
- **The compressive behavior of bone as a two-phase porous structure**
- J Bone Jt Surg, 59-A (1977), pp. 954–962
-

18.
- 78
- R. Zegarra-Parodi, P. de Chauvigny de Blot, L.D. Rickards, E. Renard
- **Cranial palpation pressures used by osteopathic students: effects of standardized protocol training**
- J Am Osteopathic Assoc, 109 (2009), pp. 79–85
- View Record in Scopus
  | Cited By in Scopus (9)

19.

- 79
- J. Marcotte, M.C. Normand, P. Black
- **Measurement of the pressure applied during motion palpation and reliability for cervical spine rotation**
- J Manipulative Med Physiol Ther, 28 (2005), pp. 591–596
- Article
  |
  PDF (231 K)
  |
  View Record in Scopus
  | Cited By in Scopus (14)

20.

- 80
- A.A. Weaver, K.L. Loftis, J.C. Tan, S.M. Duma, J.D. Stitzel
- **CT based three-dimensional measurement of orbit and eye anthropometry**
- Invest Ophthalmol Vis Sci, 51 (2010), pp. 4892–4897
- View Record in Scopus
  |
  Full Text via CrossRef
  | Cited By in Scopus (13)

1.

- 81
- D.M. Moorcroft, J.D. Stitzel, G.G. Duma, S.M. Duma
- **Computation model of the pregnant occupant: predicting the risk of injury in automobile crashes**
- Am J Obstet Gynecol, 189 (2003), pp. 540–544
- Article
  |
  PDF (346 K)
  |
  View Record in Scopus
  | Cited By in Scopus (40)

2.

- 82

- D.M. Moorcroft, J.D. Stitzel, S.M. Duma, G.G. Duma
- **The effects of uterine ligaments on the fetal injury risk in frontal automobile crashes**
- J Automobile Eng, 217 (2003), pp. 1049–1055
- View Record in Scopus
  |
  Full Text via CrossRef
  | Cited By in Scopus (10)

3.
- 83
- F.S. Gayzik, O. Bostrom, P. Ortenwall, S.M. Duma, J.D. Stitzel
- **An experimental and computational study of blunt carotid artery injury**
- Ann Adv Automotive Med, 50 (2006), pp. 13–32
- View Record in Scopus
  | Cited By in Scopus (5)

4.
- 84
- F.S. Gayzik, M.R. Shayn, H.C. Gabler, J.J. Hoth, S.M. Duma, J.W. Meredith *et al.*
- **Characterization of crash induced thoracic loading resulting in pulmonary contusion**
- J Trauma, 66 (2009), pp. 840–849
- View Record in Scopus
  |
  Full Text via CrossRef
  | Cited By in Scopus (7)

5.
- 85
- F.S. Gayzik, D.P. Moreno, C.P. Geer, D.S. Wuertzer, R.S. Martin, J.D. Stitzel
- **The development of full body geometrical data for finite element models: a multi-modality approach**
- Ann Biomed Eng, 39 (2011), pp. 2568–2583
- View Record in Scopus
  |
  Full Text via CrossRef
  | Cited By in Scopus (9)

6.
- 86

- F.S. Gayzik, J.J. Hoth, J.D. Stitzel
- **Finite element-based injury metrics for pulmonary contusion via concurrent model optimization**
- Biomech Model Mechanobiology, 10 (2011), pp. 505–520
-

7.
- 87
- J.D. Stitzel, F.S. Gayzik, J.J. Hoth, J. Mercier, H.D. Gage, K.A. Morton *et al.*
- **Development of a finite element-based injury metric for pulmonary contusion part I: model development and validation**
- Stapp Car Crash J, 49 (2005), pp. 271–289
-

8.
- 88
- J.D. Stitzel, J.T. Baretta, S.M. Duma
- **Predicting fracture due to blunt impact: a sensitivity analysis of the effects of altering failure strain of human cortical rib bone**
- Int J Crashworthiness, 9 (2004), pp. 633–642
-

9.
- 89
- F. Turquier, H.S. Kang, X. Trosseille, R. Willinger, F. Lavaste, C. Tarriere *et al.*
- **Validation study of a 3D finite element head model against experimental data**
- Soc Automotive Engineers (1996)
-

10.
- 90
- L. Zhang, K.H. Yang, R. Dwarampudi, K. Omori, T. Li, K. Chang *et al.*
- **Recent advances in brain injury research: a new head model development and validation**
- Stapp Car Crash J, 45 (2001), pp. 375–400
- View Record in Scopus

11.
- 91
- S.C. Jasinoski, B.D. Reddy, K.K. Louw, A. Chinsamy
- **Mechanics of cranial sutures using the finite element method**

- J Biomech, 43 (2010), pp. 3104–3111 [Epub 2010 Sep. 9]
- Article
  |
  PDF (1246 K)
  |
  View Record in Scopus
  | Cited By in Scopus (12)

12.
- 92
- Q. Wang, S.A. Wood, I.R. Grosse, C.F. Ross, U. Zapata, C.D. Byron *et al.*
- **The role of the sutures in biomechanical dynamic simulation of a macaque cranial finite element model: implications for the evolution of craniofacial form**
- Anat Rec (Hoboken), 295 (2012), pp. 278–288 http://dx.doi.org.ezproxy.lib.usf.edu/10.1002/ar.21532 [Epub 2011 Dec 20]
- View Record in Scopus
  |
  Full Text via CrossRef
  | Cited By in Scopus (3)

13.
- 93
- Z. Zong, H.P. Lee, C. Lu
- **A three-dimensional human head finite element model and power flow in a human head subject to impact loading**
- J Biomech, 39 (2004), pp. 284–292
-

14.
- 94
- Cormier JM. Epidemiology and biomechanical analysis of facial fractures. PhD Dissertation. Virginia Polytechnic Institute and State University. 2009.
-

Corresponding author. Tel.: +1 540 231 2465; fax: +1 540 231 2953.