Helmet Construction Influences Brain Strain Patterns for Events Causing Concussion in Youth Ice Hockey

David A. Koncan, Roger Zemek, MD, Michael D. Gilchrist, Thomas B. Hoshizaki

Abstract Two conventional ice hockey helmet liners were tested for head impact events documented to cause concussion in youth ice hockey (falls to the ice and boards). Impact parameters were established using real world cases of concussion in youth ice hockey to inform MADYMO simulations, which established a corridor of response representative of youth ice hockey impact events. Helmets were tested at two velocities on two surfaces (ice and boards), at a high and low velocity determined by the kinematic simulations. Helmets were evaluated based on impact kinematics and finite element metrics of maximum principal strains and cumulative strain damage measures using a scaled finite element model of the brain. The vinyl nitrile liner showed better performance at reducing rotational acceleration and velocity measures, whereas the expanded polypropylene liner performed better for reducing linear acceleration for most cases. The vinyl nitrile liner showed better performance in reducing maximum principal strain for impacts at and below 4 m/s, but showed signs of approaching the upper functional range at 4.5 m/s with increased strain compared to the expanded polypropylene liner.

Keywords Concussion, helmets, youth ice hockey, head impact, finite element modelling

I. INTRODUCTION

Helmets play an important role in ice hockey, protecting participants who sustain impacts to the head resulting from falling onto the ice and boards, from collisions with other players, or from stick and puck impacts. Ice hockey helmets are certified to standards with pass/fail criterion based on peak resultant linear acceleration [1]. While peak linear acceleration has been linked with catastrophic injuries, such as skull fracture and subdural hematoma [2-3], rotational motion is more influential in creating brain motion and strains [4-7], which has been reported to cause metabolic cascades responsible for concussive symptoms [8-9]. Concussions are of growing concern in youth ice hockey, with many diagnosed concussions each year [10-11]. In addition to the short-term symptoms of concussions, there are documented cognitive deficits and social interaction difficulties in youth recovering from concussion [12-14], as well as struggles behaviourally [15-16] with effects lasting months or more after the initial injury. With children still attending school, the repercussions of a single injury event can be quite severe for any child's development. Protective equipment should be evaluated for performance in managing not only impact forces, but the overall trauma experienced by the brain.

Ice hockey helmets are typically constructed using a relatively stiff outer shell with a compliant liner. Two commonly used liner materials are vinyl nitrile (VN) foam and expanded polypropylene (EPP). These materials have been used in many helmet models because they offer protection for repeated impacts without experiencing material failure [17]. Both liner materials have been tested based on impact kinematics using a linear impactor [18], as well as three different impact events common in professional ice hockey [19]. The liner's ability to reduce peak strain or cumulative strain in the brain has not yet been evaluated for the youth population, who play and experience the game differently from adults and professional players.

Youth ice hockey differs from adolescent and professional leagues in skill level and size of participants, with rules also adapted to fit the age group. As of 2013, Hockey Canada changed the rule book and now introduces

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body checking at the age of 13 [20], up from 11 years previously, with the aim of reducing head injuries and concussions. Since shoulder checks to the head represent the event causing the largest proportion of concussive injuries in professional ice hockey [21] and body checking represents the event causing the most concussive injuries in adolescent ice hockey [10], youth ice hockey (without body checking) represents a unique game environment where the greatest risk may lie in falling onto the ice and boards. The youngest ice hockey players are shorter, have smaller heads and less skating ability, so the impact characteristics are different from those of adults. With children starting organised hockey at age five, protection for this age group should be evaluated for impacts more characteristic of game play in addition to current standards tests.

Current standards assess performance in linear acceleration and have effectively reduced the occurrence of focal injuries in sports, such as skull fractures [17]. To protect against concussion in youth, it is important for helmets to prevent skull fractures as well as reduce overall brain trauma as much as possible. Metrics such as maximum principal strain (MPS) and cumulative strain damage measure (CSDM) have been used in several studies [22-25] and are indicators of the maximum and overall trauma experienced by the brain during an impact. Helmets will be evaluated using these brain strain metrics to assess performance in reducing brain trauma in addition to standard kinematic metrics.

II. METHODS

Impact conditions were chosen based on data from a Canada-wide study of youth presenting to pediatric emergency rooms. The study included the collection of biomechanical data using a standardised data-collection form, where patients between the ages of five and 18 diagnosed with a concussion as defined by the Zurich consensus statement [26] were included in the study population. Full descriptions of inclusion and exclusion criteria for the patients were described in [27].

For the present study, impact conditions were chosen based on a subset of patients; cases involving falls in youth ice hockey impacting the ice and boards were included. Complete descriptions of the impact event, including a measurement of the height of the patient, height fallen, age, sex, location of impact on the head and the impacting surface (ice, boards, collision with another player), were required from the patient, parent/guardian, or both to qualify for inclusion.

To obtain as accurate impact parameters as possible, the standardised data-collection form included a graphic separating impact locations on the head into 25 different areas where the patient, parent/guardian, or both could select one or multiple to define where exactly the impact occurred as well as whether the impact occurred from the side, upwards, or downwards. Patients could also specify if the head was the initial point of contact, or whether contact occurred with another object prior to head contact with the impacting surface. Because exact body positions could not be accounted for in this study, only cases where the impacting surface was the initial point of contact were included.

Collisions with other players were excluded as body positions and impact velocity could not be accounted for. Impact conditions were determined using the most common impact site for each impacting surface (ice, boards), at two impact velocities, establishing a corridor of performance for each helmet. In total, 66 cases of concussion were analysed to determine impact location and velocity parameters.

Mathematical dynamic models (MADYMO)

Using data from the patient intake forms containing descriptions of the impact events, simulations were conducted to establish an upper and lower boundary of head impact velocities associated with each concussive event. This software is capable of simulating human body kinematics, and has been used previously in reconstructions of head injuries resulting from falls [28-29]. For each case, a series of simulations were run with various body positions and limb positions during the fall such that the head impacted the specified location on the patient intake form. Initial body and limb positions were approximated to reflect postures typical in ice hockey. In addition to gravity, the MADYMO models were loaded with horizontal velocities ranging from 0.5 – 5.0 m/s to reflect skating velocity of youth. With cases involving youth between the ages of 5-18, 3 different pedestrian models were used in MADYMO (6-year-old, 5th percentile female, 50th percentile male). Models were chosen based on age of the patients, with males aged 5-7 and females aged 5-9 using the 6-year-old pedestrian model, males aged 8-15 and females aged 10+ using the 5th percentile female, and males aged 16-18 using the 50th percentile male.

A total of 67 concussive events fit inclusion criteria and were used to establish impact conditions, with 60 falls to the ice, and seven to the boards. For ice impacts, the most common impact site was the back of the head, with impact velocities ranging between 2.44 m/s and 4.88 m/s. For impacts to the boards, the most common impact site was the side of the head, with velocities ranging between 3.01 m/s and 3.98 m/s. Ice impacts were conducted to the rear, at 2.5 m/s and 4.5 m/s, and impacts to the boards were conducted to the side, at 3.0 m/s and 4.0 m/s.

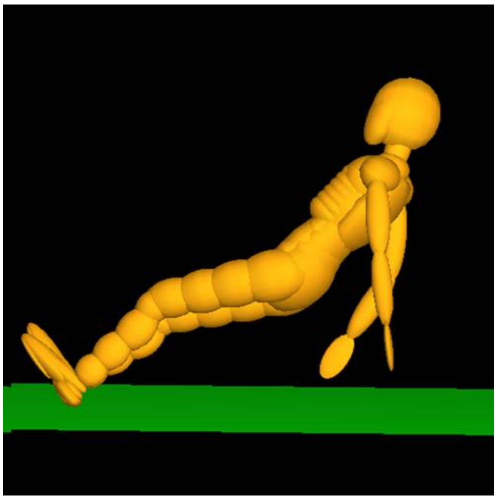

Fig. 3. Snapshot image of a MADYMO six-year-old pedestrian model mid-fall.

Experimental Testing

Two conventional helmet liner designs were tested (vinyl nitrile, expanded polypropylene) for each impact condition. Helmets were fitted onto a Hybrid III six-year-old headform, attached to an unbiased neckform. The head and neckform were attached to a carriage on a monorail drop rig, which was used for all impacts.

For impacts to the ice, an aluminum cylinder was filled with water and frozen (20 cm diameter, 9.5 cm depth). The cylinder was then rigidly attached to a concrete anvil. For impacts with the boards, a single panel of boards was constructed (4 ft x 3.5 ft, 1.22 m x 1.07 m). The boards' support structure was constructed using wooden framing, with the surface being made of a ½ inch sheet of high density polyethylene (HDPE). Test set-ups with both anvils are shown in Fig. 1 and Fig. 2.

Fig. 1. Rear impact site on the ice anvil.

Fig. 2. Side impact site with the boards anvil.

The headform used in this study was the Hybrid III six-year-old, attached to an unbiased neckform. Acceleration time histories were collected using nine Endevco 7264C-2KTZ-2-300 single-axis accelerometers (Endevco, California, USA) arranged in a 3-2-2-2 array to capture three-dimensional accelerations [30]. Data were sampled at 20 kHz, using a DTS TDAS system (DTS), and filtered using a CFC class 180 filter.

Finite Element Model of the Brain

To compute brain strains, a scaled version of the University College Dublin Brain Trauma Model (UCDBTM) was used. To reflect the size of children's heads and the headform used for testing, the model was globally scaled to

90% of the original size of the UCDBTM. This size was chosen based on MRI brain size data [31], an average for the six-year-old children involved in the study (n=9). Fit was emphasised in the anterior-posterior and inferior-superior axes (within 1 standard deviation). Details describing the finite element model and material properties are listed in the appendix. Material properties of the finite element model were not altered from the adult model following Roth et al. [32] who employed adult material properties in developing a finite element model of the brain of a 3-year-old child. While it is recognized that some studies have identified differences in brain material properties due to age [33-35], this was not addressed in this study.

The finite element (FE) model will be used to calculate the brain strain response to impacts, extracting values of MPS and CSDM. Cumulative strain damage measures will be output for the volume fraction of brain tissue in the cerebrum experiencing strain above 0.10, 0.15, 0.20 and 0.25, which will be labelled CSDM10, CSDM15, CSDM20 and CSDM25, respectively.

III. RESULTS

Helmet performance varied between liner designs across all metrics, with no one helmet consistently outperforming the other. Though statistical significance was not reached in all cases, the EPP liner mostly outperformed the VN liner for peak resultant linear acceleration. However, the VN liner mostly outperformed the EPP liner for rotational acceleration and rotational velocity. A summary of the experimental impact kinematics is shown in Table I.

TABLE I

EXPERIMENTAL IMPACT KINEMATICS RESULTS SUMMARY FOR TWO HELMET LINERS (VN, EPP) AND IMPACT ANVILS (ICE, BOARDS)

	Peak resultant linear acceleration (g)		**Peak resultant rotational acceleration (rad/s^2)**		**Peak resultant rotational velocity (rad/s)**	
Impact details	**VN**	**EPP**	**VN**	**EPP**	**VN**	**EPP**
Ice – 2.5 m/s	63.6	52.7	2647	3087	33.2*	37.7*
Ice – 4.5 m/s	145.2*	117.7*	5640	5170	41.9*	47.3*
Boards – 3 m/s	44.1	42.4	3557	3890	38.5	39.7
Boards – 4 m/s	58.5*	63.0*	4523*	5863*	45.8	49.9

*denotes statistically significant differences between helmet liners ($p<0.05$).

Finite element model simulation responses followed the trends of rotational metrics, with the VN liner outperforming the EPP liner in most conditions. Figure 4 shows simulation results of MPS for all impact conditions, with Fig. 5 and Fig. 6 detailing the simulation results of CSDM trends.

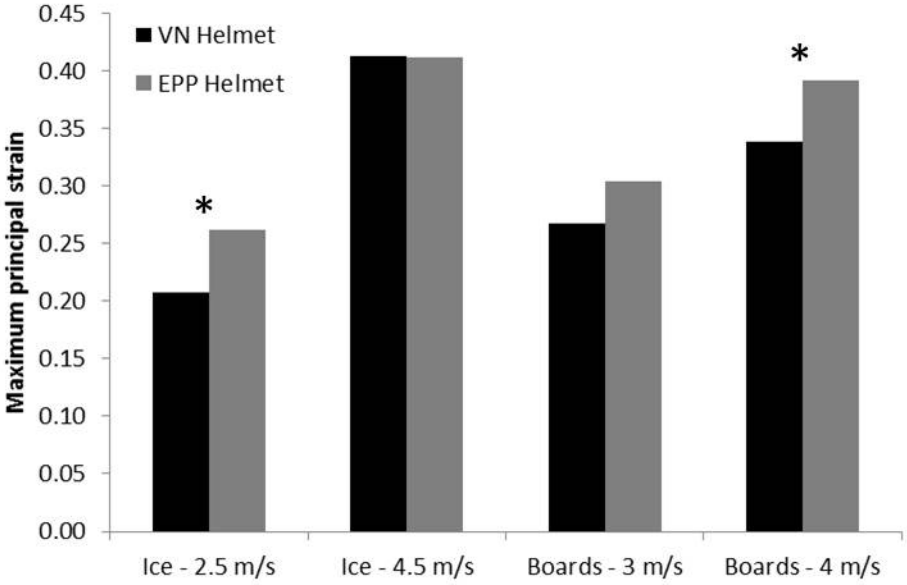

Fig. 4. Simulation results of maximum principal strain for the VN liner (black) and EPP liner (grey) across ice and boards impacts. Statistical significance between liners denoted by * (p<0.05).

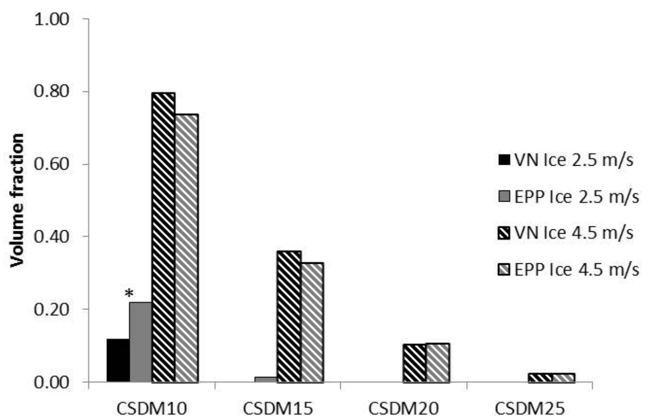

Fig. 5. Simulation results of CSDM measures for ice impacts for VN and EPP helmet liner designs, with the volume fraction of the cerebrum experiencing strain over 0.1, 0.15, 0.2 and 0.25. Statistical significance between liners denoted by * (p<0.05).

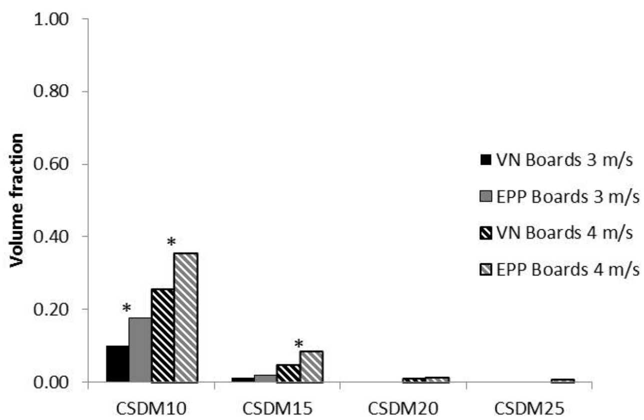

Fig. 6. Simulation results of CSDM measures for boards impacts for VN and EPP helmet liner designs, with the volume fraction of the cerebrum experiencing strain over 0.1, 0.15, 0.2 and 0.25. Statistical significance between liners denoted by * (p<0.05).

The VN liner outperformed the EPP liner for MPS results in three of four conditions, two of which were statistically significant. For CSDM measures of impacts with the boards, the VN liner outperformed the EPP liner, with three metrics reaching statistical significance (CSDM10 at 3 m/s and 4 m/s, CSDM15 at 4 m/s). For ice impacts, the VN liner outperformed the EPP liner at low velocity, but not at high velocity. Trends for MPS and CSDM results did not consistently identify either liner as superior in all cases.

IV. DISCUSSION

The impact kinematics show that the EPP liner performs better for linear acceleration compared to the VN liner but not for rotational acceleration, which is similar to the findings of a previous study of VN and EPP liners for hockey helmets [18]. Rotational velocity results show similar trends to rotational acceleration, with the exception of the ice impact at 4.5 m/s. In this case, the peak resultant rotational acceleration for the VN liner was above that of the EPP liner, but the rotational velocity was still below that of the EPP liner. This is attributed to differences in impact duration and curve shape from how impact energy is transferred for each impact anvil.

The impact characteristics chosen for this study were based on cases of concussion, so it is not surprising that some metrics correspond to risk of concussive injury in adult data. Determining risk of concussive injury in children is difficult due to a lack of published impact data relating to the age group. The data used to determine risk for concussive injury were based on adult American football data [22][36], and some scaled from animal data [37] (Table II). As it is unknown whether risk of concussion in children differs from adults, comparisons based on adult data should be made with caution. According to the adult data in Table II, risk of concussive injury would only reach or exceed 50% for peak linear acceleration for one impact condition (ice, 4.5 m/s), none would exceed 50% risk for peak rotational acceleration, and all would exceed 50% risk for rotational velocity. With all impact characteristics in this study chosen to reflect a range of concussive impacts, the discrepancy between current impact kinematics and published values of risk from adult data support two ideas: impact kinematics in adult American football differ from youth ice hockey, therefore metrics identifying risk should be investigated for youth ice hockey; and children may have different tolerances to impact forces causing concussion.

TABLE II
PUBLISHED EXPERIMENTAL IMPACT KINEMATICS CORRESPONDING TO RISK OF CONCUSSIVE INJURY IN ADULTS FROM AMERICAN FOOTBALL AND SCALED ANIMAL DATA

Risk of Concussion	Peak Resultant Linear Acceleration (g)	Peak Resultant Rotational Acceleration (rad/s^2)	Peak resultant Rotational Velocity (rad/s)	Source
50%	-	1800	20 - 30	Scaled animal data [37]
50%	82	5900	-	American football [22]
50%	-	6383	28.3	American football [36]

The effect of neck stiffness was not addressed in this study. While it is acknowledged that neck stiffness can influence impact kinematics, it has been shown that in adults, changes in neck stiffness of 30% have limited effect on impact kinematics and strains in the brain [38]. Additionally, in a study of collisions in youth ice hockey it was found that there was no statistical difference in head kinematics resulting from anticipated and unanticipated collisions [39]. It is inferred that anticipated collisions would have increased neck stiffness where the player would brace for the collision compared to unanticipated collisions, with no resulting differences in impact kinematics. As a result, neck stiffness was not investigated in this study. A hair surrogate was not used in these tests, which could possibly influence kinematics by allowing more sliding between the helmet and headform. Head-helmet interfaces have been investigated using the Hybrid III adult 50th percentile headform, finding no significant differences in response between a bare headform and three different surface conditions (nylon, wig, oil bladders) [40]. Based on these findings and due to the centric nature of the impacts conducted in this study, it is assumed that slipping at the helmet-head interface is minimal.

Patterns of brain strain were affected by helmet liners, highlighted by the trends in CSDM measures. Different levels of force were transmitted to the brain, resulting in changes in volume fractions experiencing each measured level of strain. The VN liner showed better performance in CSDM measures for all impacts with the boards, with less brain tissue experiencing strain overall, as well as a lower MPS. For ice impacts, the VN liner outperformed the EPP liner at 2.5 m/s, with lower MPS and lower CSDM values, however at 4.5 m/s the MPS values were similar and the EPP liner outperformed or was similar to the VN liner at each level. The VN liner effectively manages impact forces being transmitted to the brain at low velocity. At high velocity, however, the liner is losing the ability to effectively manage impact forces, evidenced by the higher linear acceleration. The ice impact at 4.5 m/s is the fastest and least compliant impact tested in this study, and may be approaching the upper functional range of this particular VN foam liner. Each type of liner can be tuned by manufacturers, allowing for different compliances for each type of foam. The compliance and performance of the two liners tested in this study are likely related to hockey helmet certification standards and helmet designers' choices for protection. Ice hockey standards test near 4.5 m/s on minimally compliant surfaces. The liners that were tested show evidence of being designed for protection at different impact velocities, with the VN being more compliant and offering protection velocities below and up to impact standards. The EPP liner is stiffer and offers more protection at velocities close to impact standards, and likely higher as well.

In order to better protect youth ice hockey players, helmets should be developed to limit force transmission to the brain. The VN foam liner performed better than the EPP in certain conditions, likely where the VN was able to deform and shear, managing both linear and rotational forces during impact. The EPP liner does not have the same ability to shear under load compared to VN foam, and as a result the rotational metrics were, in general, higher than the VN. Since brain tissue is vulnerable to shear induced by rotational motions [6][41-43], helmet designs should incorporate technology to help manage rotational energy in order to reduce strains in the brain tissue.

Limitations

This study relied on patient intake forms to establish the impact parameters of each concussive case, which were then used to establish impact conditions for each impacting surface. Strict inclusion criteria were used in order to reduce variance in establishing impact conditions due to interpretations of form data, eliminating any cases with missing, confusing, or conflicting entries. Despite these strict criteria, the MADYMO kinematic simulations are limited in describing exact injury events, and are used instead to establish a range of possible impact velocities for the event described on patient intake forms, giving both a conservative and liberal estimation of how concussions can occur in youth ice hockey falls. Additionally, while it is recognized that there are studies that have reported age-dependent material properties of the brain, this study did not address this issue. If material properties were applied more similar to those used in pediatric models of the brain [44-45], it would be likely that strain values would increase as the properties are more compliant than those used in this study.

V. Conclusions

The VN and EPP ice hockey helmet liners were shown to produce different strain responses in the brain resulting from impacts to the ice and boards. Strain distribution measures suggest that for impacts at 4.5 m/s to the ice, the VN liner is approaching the upper functional limit of the material evidenced by increased CSDM compared to the EPP liner. Both strain metrics suggest that for impacts below 4.5 m/s, the VN liner shows slightly better protection to limit both peak strain and cumulative strain. Above 4.5 m/s, the current VN liner material may not be adequate, and a different foam material with higher stiffness would be required to effectively manage impact forces at the higher velocities if protection at higher impact velocity is desired.

Impact kinematics of the youth helmets followed previous tests of adult ice hockey helmets, which showed that an EPP liner performs better for reducing peak linear acceleration, with VN performing better for reducing rotational acceleration and velocity. Of the kinematic metrics, rotational velocity consistently identified over 50% risk of concussive injury in each case compared to the adult and scaled animal data. Rotational velocity may be a suitable metric to investigate for use in studies involving children to compare to adult data or to scaled animal data.

VI. Acknowledgement

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VIII. APPENDIX

The UCDBTM geometry was based on a male cadaver using medical imaging techniques [46-47]. The model includes the scalp, skull, dura, CSF, pia, falx, tentorium, grey and white matter, cerebellum, and the brain stem. Validation tests were conducted against cadaveric data of intracranial pressure from [48], as well as brain motion using neutral density targets from [4].

Material properties of the tissues represented in the model are shown in Table III. The brain tissue was modelled with a combination of hyperelastic and viscoelastic properties. The shear characteristics of the brain were expressed by:

$$G(t) = G_\infty + (G_0 - G_\infty)e^{-\beta t} \tag{1}$$

where G_∞ represents the long-term shear modulus (Pa), G_0 the short-term modulus (Pa), and β the decay factor (s^{-1}). The Mooney-Rivlin hyperelastic material model was also used, given by:

$$C_{10}(t) = 0.9C_{01}(t) = 620.5 + 1930e^{-t/0.008} + 1103e^{-t/0.15}, (Pa) \tag{2}$$

where C_{10} and C_{01} are the mechanical energy absorbed by the material when the first and second strain invariant change by a unit step input, respectively [49-50], and t is the time (s).

TABLE III
MATERIAL PROPERTIES FOR THE UCDBTM [46-47]

Material	Young's modulus (MPa)	Poisson's ratio	Density (kg/m^3)
Facial bone	500	0.23	2100
Cortical bone	15000	0.22	2000
Trabecular bone	1000	0.24	1300
Scalp	16.7	0.42	1000
Dura	31.5	0.45	1130
Pia	11.5	0.45	1130
Falx and tentorium	31.5	0.45	1130
CSF	15	0.5	1000

	Shear modulus (kPa)		Decay constant (s^{-1})	Bulk Modulus (GPa)	Density (kg/m^3)
	G_∞	G_0			
Grey matter	2	10	80	2.19	1060
White matter	2.5	12.5	80	2.19	1060
Cerebellum	2	10	80	2.19	1060
Brain stem	4.5	22.5	80	2.19	1060