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American football helmet for preventing concussion, a literature review

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Abstract

This paper reviews the studies that have been conducted on the performance of the American football helmet in preventing concussion. The review will also guide us to understand what problems still exist and what research directions we should take. Throughout the history of sports, injuries limiting the career life of the athletes have been the leading concern of the sport authorities. These injuries are more extensive in sports in which athletes are in severe contact with each. Mild Traumatic Brain Injury, concussion, widely occurs in American football because of the frequent strokes to players' heads. Concussion includes several types ofneurological dysfunctions such as headache, dizziness, confusion, blurred vision, delayed reaction time and etc.Lots of the studies have focused on understanding the concussion and improving the protective performance of the helmet, so that the dose of the injury in players is reduced. Researches in this area can be classified as two major methods: experimental studies and Finite Element Modeling (FEM) simulations. In experiments, researchers have tried torecord head impacts or reconstruct the severe collisions using the game videos in the laboratory conditions. They have used the Hybrid III dummy in order to study the effects of the different impact parameters such as direction, velocity, region of the head being hit and etc. These studies have been done by analyzing the dynamic responses of head includinglinear acceleration (LA), rotational acceleration (RA), and different head injury criteria like Head Injury Criteria (HIC) and Gadd Severity Index (GSI).Mentioned impact parameters have been also examined using FEM simulations. Researchers have applied the results of experimental tests including linear and rotational acceleration in order to study the brain deformation responses to different types of impacts. In this regards, brain deformation responses like maximum principal strain have been considered and analyzed using the head injury and concussion thresholds.

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Keywords: American football; Concussion; Linear acceleration; Rotational acceleration; Finite element modeling

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1. Introduction

Concussion is a form of mild traumatic brain injury which is caused due to the accelerating movement of the head [1]. A recent research has shown that 1.6 to 3.8 million of sport related traumatic brain injuries occurs annually in USA [2], with 300,000 being sport related concussions [3,4, 5]. It is obvious that this large number of injuries cause a great amount of annual medical costs. In addition to the medical costs, these injuries result in long-term physical and psychological problems in injured players. The long-term effects of concussion and its economical outcomes have attracted a lot of researchers to study different aspects of this injury in order to gain in-depth understanding of it. Early studies about concussion concentrated on animal experiments [6,7, 8]. These animal studies helped to obtain initial understanding about the injury while being difficult to apply for humans [9]. Nowadays, these studies have progressed to measuring the head impacts in real time using accelerometers. Further studies have been conducted using FEM simulation of the brain and its tissue deformation due to impact.

2. Experimental studies using real-time data

A great part of efforts for evaluating the performance of American football helmet in preventing concussion have focused on analyzing the head impact data recorded during the real-time games and trainings. Some other researchers have tried to exactly reconstruct the head impacts based on the recorded videos from the National Football League (NFL) using Hybrid III dummies and test setups in laboratory scale. The first measurable responsesof the head are linear and rotational accelerations which reflect the performance of the helmet. For injury cases,concussions have been caused due to long range of accelerations, with peak linear accelerations ranging from 48g to 188g [10], and peakrotational accelerations ranging from 2174 rad/s2 to 9678 rad/s2. In addition to these responses, scientists have developed different criteria, like Head Impact Power (HIP, rate of change of kinetic energy), head impact jerk (HIJ, rate of change of head input acceleration), Gadd Severity Index (SI) and Head Injury Criteria (HIC). Using these measures along with other biomechanical aspects of the head impact including location and direction of the impact and position of the impacted player, researchers have attempted to study concussion injury. The results of these studies can be classified as magnitudes of the non-injury and concussive impacts, concussion thresholds, impact location and position of the impacted player.

2.1. *Magnitudes of dynamic responses in non-injury and concussion cases*

Several researchers recorded the head impacts during the training or/and game sessions of the high school and collegiate teams [11,12,13,14,15,16,17,18,19,20]. Some other researchers measured the head dynamic responses from the laboratory reconstruction of the NFL game impacts [21, 22]. The summary of the data that have been recorded in various researches for non-injury and concussion casesare listed in Tables 1 and 2.

Table 1. Average of head peak dynamic responses for non-injury cases.[1]All impacts including a few injuries

	LA (g)	RA (rad/s2)	GSI	HIC
Naunheim et al, [11]	29.2±1.1	-	38.4±6.6	22.5±3.6
Duma et al, [12]	32±25	2213.56	36±91	26±64
Brolinson et al, [16]	20.1±18.7		-	-
Broglio et al, [13]Game	24.76±15.72	1669.79±1249.41	-	-
Training	23.26±14.48	1468.58±1055.00	-	
Rowson et al, [15]	22.3	1335	-	-
Broglio et al, [14]	25.1±15.4[1]	1627.1±1182.9[1]	-	-
Rowson et al, [17]	-	1230±915	-	-
Zhang et al, [21]	55	3938	-	-
Pellman et al, [22]	60±24	4235±1716	154±82	121±64

Based on the data in Table 1, it turns out that all researchers except two[21,22] recorded relatively similar and low data for non-injury cases. But, the averaged magnitudes measured by Zhang et al. [21] and Pellman et al. [22] for non-injury cases have been significantly higher, because they biasedly selected the most severe impact cases to study in laboratory scale, while others have chosen all impacts during the games and trainings.

Table 2. Average and range of head peak dynamic responses for concussion cases.

	#concussion	LA (g) Ave (Range)	RA ($\mathbf{rad/s^2}$) Ave (range)	GSI Ave	HIC Ave
Duma et al. [12]	1	81 (-)	7912 (-)	26	200
Brolinson et al. [16]	3	103.3 (55.7-136.7)	-	-	-
Schnebel et al. [18]	6	- (81.9-145.7)	-	-	-
Guskiewicz et al. [19]	13	102.8 (60.51-168.71)	-	-	-
Zhang et al. [21]	22	94 (48-138)	6398 (2615-9678)	-	-
Pellman et al. [22]	25	98±28 (48-138)	6432±1813 (2615-9678)	474±252	381±197
Broglio et al. [14]	13	105.0 (74.0-146.0)	7229.5 (5582.6-9515.6)	-	-
McAllister et al. [20]	9	74.9±22 (40.6-11.6)	4760±1350 (2174-6325)		
Rowson et al. [17]	57	-	5022±1791 (-)	-	-

2.2. *Dynamic based concussion thresholds*

Using the experimental data, researchers have calculated thresholds, which can predict the occurrence of the concussion. These thresholds have been set for peak linear and rotational acceleration, rotational velocity (RV) and brain injury criteria like SI and HIC. Thresholds for different probabilities of injury are summarized in Table 3.

Table 3. Thresholds based on the dynamic responses for different probabilities.

	Probability	LA (g)	RA ($\mathbf{rad/s^2}$)	RV ($\mathbf{rad/s}$)	GSI	HIC	HIP
Newman et al. [23]	50%	77	6322	-	291.2	239.8	12.79
	95%	115	9267	-	558.9	485.2	20.88
Pellman et al. [22]	Nominal	-	-	-	300	250	-
King et al. [24]	25%	57	4384	-	-	136	-
	50%	79.3	5757	-	-	235	-
	75%	98.4	7130	-	-	333	-
Zhang et al. [25]	25%	66	4600	-	-	151	-
	50%	82	5900	-	-	240	-
	80%	106	7900	-	-	369	-
Broglio et al. [14]	Nominal	96.1	5582.3	-	-	-	-
Rowson et al. [17]	50%	-	6383	28.3	-	-	-
	75%	-	6945	30.8	-	-	-

2.3. *Impact location*

Another biomechanical aspect of concussion that has been widely studied by researchers is the location and direction of the impacts based on three factors: frequency of impact, severity of impact and concussive impact. The purpose of these studies has been to find the regions of head being impacted more frequently and more severely, and also regions of the head that have been more vulnerable to concussion under the effect of impacts.

Regarding the frequency of the impacts,several research results have shown that front of the helmet receives impacts more frequently than other regions of the head. Rowson et al. [15] showed that impacts to front of the helmet were the most common followed by left, right, back and top regions. Rowson et al.[17] classified their recorded impacts to three cases including impact to front and back of the helmet, impacts to sides of the helmet and impact to top of the helmet. Based on their results, most of the impacts (67.5%) were to the front and back of the helmet, followed by sides and top of the helmet. Crisco et al.[26,27]suggested that impacts to front and top of the helmet are respectively the highest and least frequent impacts. Greenwald et al. [28]showed that impacts to front of the head were more frequent (43.1%) than back (24.4%), sides (19.5%) and top (13.0%) regions. Broglio et al.[13] stated that front of the helmet would receive more frequent impacts than back, side and top regions.

In order to analyze the severity of the impacts, Mihalik et al.[30] showed that players are at least 6.5 times more likely to experience impacts with peak linear acceleration of greater than 80g to the top of the head than to the back, front, left and right sides of the head. Rowson et al.[17] concluded that impacts to top of the helmet had caused lower magnitudes of peak rotational acceleration than impacts to front and back of the helmet. Crisco et al.[27] showed that impacts to top of the helmet are associated with experiencing the lowest peak rotational acceleration and the highest peak linear acceleration. Broglio et al. [13] revealed that impacts to the top of the head produced the greatest amount of peak linear acceleration and force, followed by front, back and sides. They also revealed that impacts to front of the head had resulted in the greatest peak rotational acceleration, followed by back, sides and top. Similar to all of these researches, Pellman et al. [22,29] showed that the facemask impacts had the lowest average of peak linear acceleration and relatively high average of peak rotational acceleration.

Regarding the concussive impact, some researches demonstrated that brain is more vulnerable to sustain concussion due to the impacts to top of the head. Mihalik et al. [30] stated that four out of seven concussion cases were due to the impacts to top of the head. Similarly, Guskiewicz et al. [19] stated that 6 out of 13 concussion cases occurred as the result of impacts to top of the head. However, several researches have proposed front region of the head being more susceptible to concussion. Greenwald et al. [28] recorded 17 concussion cases, involving eight impacts to the front, three to the top, five to the sides and one to the back of the helmet. Broglio et al. [14] recorded 13 concussive impacts in which 8 concussion cases were due to the impacts to front of the helmet. Between 25 concussion cases recorded by Pellman et al. [22, 29], 14 cases involved loading on the facemask. Based on their results, the average peak linear acceleration for concussion caused by impacts to the facemask was 78±18g, while this average for other regions of the helmet was 107 to 117g. Most of the concussion cases (33 out of 57) recorded by Rowson et al. [17] were due to the impacts to front and back of the helmet, followed by impacts to top of the helmet (17 out of 57). However, the number of concussions per impact has been the most for top of the helmet.

2.4. *Player position*

Several studies have attempted to analyze different football positions based on the frequency and severity of the head impacts. In summary, they have obtained similar results stating that defensive and offensive linemen tend to receive more frequent impacts[26, 27, 30, 13, 12, 18, 32], but with lower severity[26, 27, 13, 32, 18]. In contrast, skill players (non-linemen) such as running backs, quarter backs and wide receivers experience less frequent head impacts[26, 27,30, 12, 18, 32] while causing greater severity[26, 27, 30, 18, 32].

3. FEM simulations

Several studies suggested that it is the brain deformation responses that dictate the injury, not head dynamic responses. Therefore, it is important to study how the brain tissues are affected and deformed due to the head impacts. Researchers have used head dynamic responses including linear and rotational acceleration as the input to different finite element models of the brain to study brain deformation responses like strain and stress.

Zhang et al. [21, 25], King et al. [24] and Viano et al. [35] studied brain deformation responses using data from laboratory reconstruction of NFL game videos [33] as the input to the Wayne State University Head Injury Model (WSUHIM) [34]. Using another finite element model of the head [36, 37], Kleiven[38] compared several head dynamic based metrics along with brain deformation responses from recorded concussion cases. McAllister et al. [20] studiedstrain and strain rate at corpus callosum as predictor of concussionusing Dartmouth Subject-Specific FE Head Model. Post et al. [39,31] applied a centric (through head center of gravity) and non-centric impact condition on the front, back and sides of helmeted Hybrid III dummies to study the relation between the head dynamic and brain deformation responses. They impacted the helmeted Hybrid III dummies at nine centric and non-centric sites (at 7.5 m/s) and used the dynamic responses as the input to the University College Dublin Brain Trauma Model (UCDBTM) [40, 41]. Post et al. [42, 43] studied the performance of the American football helmet being impacted with three different velocities (5.5, 7.5 and 9.5 m/s) at twolocations (one centric and one non-centric) based on head dynamics and brain deformation responses.

3.1. *Brain deformation responses*

Zhang et al. [21] and King et al. [24] demonstrated that, for injury cases, the concentration of high maximum principle strains (S) were in white matterof frontal lobe and central core region of the brain, more specifically in the midbrain, upper brain stem and most of the diencephalon. But, corpus callosum did not experience high strains. The results for strain rate (SR) and product of strain and strain rate (SSR) showed their concentration in the midbrain. Zhang et al.[25]also showed that, for intracranial pressure (P), the impact initially caused positive pressures at the impact site and then the pressure gradient progressed to the opposite side of the brain with negative pressures. Their result for shear stress (SS) showed that initially it was initially high at cortical surface of the brain and graduallymoved to the central core region of the brain. Midbrain has experienced the highest magnitudes of shear stress during the impact, followed by thalamus. The corpus callosum region, being reported to commonly experiences diffuse axonal injury, did not experience high shear stresses. Viano et al. [35] showed that during the early time after impact, high strains and strain rates concentrated in the temporal lobe close to the impact site. During the mid-time response after the impact, these concentrations moved to the temporal lobe in the far side of the brain. Finally, during the late-time response, they concentrated at the midbrain regions including fornix and Ammon. Approximately, half of the cases showed this migration of high responses. All of the concussion cases, showed the late high strain responses in regions of the fornix, midbrain and corpus callosum. Based on the results by Kleiven[38] for a concussed player, the concentration of high strains was seen in the corpus callosum, left temporal lobe and right superior part of the cortex. On the other hand, the high strain rate and von Mises stress concentrated at brainstem and midbrain. Generally, pressure linearly changed from maximum positive magnitudes close to the impact location to maximum negative magnitudes at opposite side of the impact. The FE simulations for 10 cases of concussion by McAllister et al. [20] showed the regions of high maximum principle strain in and around the corpus callosum. Post et al. [31] showed that, for most of the impact sites, dorsolateral prefrontal area and visual cortex experienced the greatest and lowest amounts of peak von Mises stress, respectively. Primary motor cortex and primary somatosensory cortex sustained the largest maximum principle strains. The same as the von Mises stress (VMS), visual cortex received the lowest maximum principle strain for most of the impact sites. Similar to other researchers, regions that sustained the larger amount of deformation were generally at opposite side of the impact side. Table 4 shows the magnitudes the brain deformation metrics in different regions of the brain.

Table 4. Brain deformation responses at different regions of the brain (T: Thalamus, M: Midbrain, CS: Coup site, CCS: countercoup site, CC: corpus callosum). [1]Midbrain, Thalamus, hypothal, fornix, ammon, parahipp and orbito-frontal-temporal. [2]Midbrain, upper brainstem and diencephalon.

	S	SR (s^{-1})	SSR (s^{-1})	SS (kPa)	VMS (kPa)	P (kPa)
Zhang et al. [21]	-	23-140 (84) [2]	Midbrain 36	-	-	-
Zhang et al. [25]	-	-	-	T: 4.5±1.2 M: 8.4±2.2	-	CS: 90±24 CCS: 76±25
Viano et al. [35]	0.317 to 0.448[1]	61.4 to 81.5[1]	-	-	-	-
Kleiven. [38]	-	-	-	-	M: 56±35	-
McAllister et al. [20]	CC: 0.28±0.9	CC: 55.6±35.7	-	-	-	-

3.2. *Brain deformation based concussion thresholds*

As it was mentioned before, concussion injury is caused due to the deformation in brain tissues. Thus, it has been proposed that thresholds based on the brain deformation responses are more successful in predicting the concussion. Table 5 summarizes the brain deformation based thresholds that have been set for different regions of the brain.

Table 5. Brain deformation based thresholds for different brain regions.

	Probability	Brain region	S	SR (s^{-1})	SSR (s^{-1})	SS (kPa)	VMS (kPa)	P (kPa)
Zhang et al. [21]	25%	Midbrain	0.25	46	14	-	-	-
	50%		0.37	60	19	-	-	-
	75%		0.49	80	24	-	-	-
King et al. [24]	25%	Midbrain	-	46	14	-	-	-
	50%		-	60	19	-	-	-
	75%		-	80	24	-	-	-
Zhang et al. [25]	25%	Midbrain	0.14	-	-	6.0	-	-
	50%		0.19	-	-	7.8	-	-
	80%		0.24	-	-	10.0	-	-
Kleiven[38]	50%	Grey matter	0.26	48.5	10.1	-	-	+68.5
	50%	White matter	-	-	-	-	-	-55.1
	50%	Corpus callosum	0.21	-	-	-	8.4	-

4. Discussion

4.1. *Concussion thresholds*

Although, dynamic based thresholds have been successful in predicting several concussions,a large number of recorded impacts in previous studies[12, 13, 16, 28, 15] are located above these thresholds, and only a few or no concussive impacts have happened.This inconsistency may have happened because major number of players experiencing concussion did not report their injury because of not understanding the concussion symptoms [12, 44,45]. But, this fact cannot completely justify the inability of the dynamic based thresholds in accurately predicting the injury.There are impacts which are below the thresholds, but they have caused concussion in players. For example, five out of thirteen concussion cases in a study [19] had lower peak linear accelerations than the thresholds set by King et al [24].Most of the dynamic responses used in a study[31] as the input to the FEM simulations were below thethresholds proposed in other researches[23, 24, 25] for 50% probability of concussion, but the results for brain deformation responses predicted high probability of concussion based on the brain deformation thresholds [25,47,38].Observing these inconsistencies in different researches imply that more than the dynamic responses should be used in setting concussion thresholds and standards, and there exist more effective parameters contributing to the incidence of concussion. These parameters might include concussion history and frequency of non-injury impacts sustained by the player before the concussion (sub-concussive impacts)[46, 19].Thus, it has been proposed that thresholds based on the brain deformation responsescan help to predict concussion more accurately.

4.2. *Location of the impact*

Regarding the severity of impacts, generally, researches demonstrated that impacts to top of the helmet resulted in the greatest and lowest amounts of peak linear acceleration and rotational accelerations, respectively. In contrast, impacts to front of the helmet caused the greatest amounts of peak rotational acceleration. Post et al. [31] supported these results, where they impacted the front, back and sides of the helmeted hybrid III dummies using a linear impactor at same speed. Their results showed that two of the impacts to front of the helmet caused the greatest peak rotational accelerations and relatively lower peak linear accelerations. Having the larger amounts of peak rotational acceleration due to the impacts to front of the helmet has been suggested to happen because of the large moment arm resulting from facemask’s distance from the head center of gravity [15] and stiffness of the facemask [31].

Regarding the susceptibility of different brain regions to concussion, there are inconsistencies between the results of the previous studies. A few researchers have proposed that impacts to the top of the head would be more concussive, and more researchers have shown that front of the head is more susceptible to concussion. Based on previous researches [22, 29], Greenwald et al. [28] stated that impacts to the helmet facemask would cause concussion with lower linear accelerations than impacts to other regions of the helmet shell. They suggested that this might be due to stiffness of the facemask, interactions between facemask and helmet shell, and physical properties of the head and neck. Besides, the statistical analysis by Viano et al. [35] showed that frontal oblique impacts caused higher mid-late strain and strain rates in the midbrain and fornix than other impact sites. So, it can be claimed that impact to front of the helmet are more concussive.

4.3. *Head dynamic responses vs brain deformation responses in brain regions*

In summary, researchers found various regions of the brain including frontal lobe, midbrain, diencephalon, fornix, left temporal lobe, right superior part of the cortex, dorsolateral prefrontal area, primary motor cortex and primary somatosensory cortex associated with concentration of peak magnitudes of strain and strain rate resulting from concussion. However, for the shear and von Mises stress, studies demonstrated less inconsistency as high magnitudes of stress concentrated at the midbrain. The inconsistency for strain shows its sensitivity to the magnitude and direction of accelerations and impact location. Post el al. [42, 31,43] demonstrated that increase in the impact velocity not only changed the magnitudes of high strains, but also changed the region of the head experiencing these high strain concentrations. On the other hand their results for von Mises stress showed less sensitivity to the impact characteristics, where increase in the impact velocity increased the magnitudes of peak von Mises stress, but it did not change the brain region sustaining high stresses. However, both von Mises stress and maximum principle strain were sensitive to the impact location. In general, analyses by Post et al.[31] showed that the brain region, which experiences the maximum amount of deformations, varies depending on the impact location. It should be noted that a part of these differences, have been because of the fact that researchers did not conduct the simulations with the same material properties of the brain. Besides, their finite element models did not include the same brain regions.

Statistical analyseshave shown that strain and stress are significantly correlated with rotational acceleration, and not significantly correlated with linear acceleration[25,35,38]. More specifically, Post et al.[39] showed that, for the centric impacts, both linear and rotational accelerationsare significantly correlated with the maximum principal strain. However, for the non-centric impacts, just rotational acceleration is significantly correlated with strain.Considering these facts, it turns out that not only both linear and rotational acceleration contribute to the brain deformation and concussion, but also other parameters such as impact location, impact direction and being centric or non-centric affect the severity of brain tissue deformations.

Theresults for pressure gradient have been similar between the previous studies. Head impacts generally cause high pressure gradients near the impact site during the early time after the impact. During the late time, the high pressure gradients move to the opposite side of the brain and concentrate at locations far from the impact site.Statistical analyses have demonstrated that pressure gradients inside the brain are significantly correlated with the magnitude and direction of the linear acceleration[25,38], which means that increase in the linear acceleration increases the magnitude of pressure, and pressure gradient moves in the direction of the linear acceleration.

5. Conclusion

In summary, the following conclusions can be made based on the previous studies.

- Front region of the helmet receives the most frequent impacts during American football games.
- Impacts to top of the helmet cause the greatest and lowest amounts of peak linear and rotational accelerations, respectively. Impact to front of the helmet causes the greatest amount of peak rotational acceleration.
- Among all regions of the helmet, more concussions have been recorded due to the impacts to front and facemask of the helmet. This may imply that front of the head is more critical inexperiencing concussion.
- Among the players' positions,defensive and offensive linemen receive more frequent impacts, but with lower severity. In contrast, skill players (non-linemen)experience less frequent head impacts, but with greater severity.
- Thresholds based on dynamic responses are not successful in accurately predicting concussion for all players.
- In addition to linear and rotational accelerations, other parameters contribute tooccurrence of concussion, parameters such as frequency of sub-concussive impacts and concussion history.
- Peak deformation responses and brain region experiencing these peak deformationswere different in previous researches, becausethey depend on different characteristics of the impact including impact location, severity and direction.
- Strain and stress in brain tissues are significantly correlated with rotational acceleration, while intracranial pressure is significantlycorrelated with linear acceleration.

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