Durability of Ice Hockey Helmets to Repeated Impacts

By

Roghieh Hakim-Zadeh

Master of Arts

Department of Kinesiology & Physical Education

Faculty of Graduate Studies, and
Faculty of Education
McGill University
Montreal, Quebec, Canada

June 2002

Your file Votre référence
ISBN: 0-612-85856-1
Our file Notre référence
ISBN: 0-612-85856-1

Abstract

This study evaluated the mechanical durability of ice hockey helmets for multiple impacts at defined energy levels. A monorail drop testing apparatus was used to conduct controlled impact tests according to the CSA standard (CAN/CSA-Z262.1-M90). Five ice hockey helmet models were tested, for a total sample of 45 helmets. All helmets were impacted up to 50 times at each of in four different locations (i.e. front, right side, back, and crown), at one of 40, 50 or 60 J of kinetic energies. In general, by increasing the impact energy, the impact acceleration attenuation properties of the helmets was decreased significantly (from 4% to 80%). Although all the helmets meet the CSA standards, attenuation properties were found to be substantially reduced beyond three repeated impacts and above 40 J impact energy. In particular, all helmets showed effective multiple impact attenuation properties at the crown, front, and rear sites; however, poor multiple impact attenuation durability was evident at the side.

Résume

Cette étude a évalué la longévité mécanique des casques d'hockey de glace pour des impacts multiples aux forces définies. Un appareillage d'essai de baisse de monorail a été utilisé pour effectuer les essais commandés au choc selon la norme de CSA (Can/csa-z262.1-m90). Cinq modèles de casque d'hockey de glace ont été examinés, pour un échantillon total de 45 casques. Tous les casques ont été effectués jusqu'à 50 fois à chacune de dedans quatre côtés différents d'endroits (c.-à-d. avant, droit, dos, et couronnes), à un de 40, de 50 ou de 60 J d'énergies cinétiques. En général, en augmentant l'énergie d'impact, les propriétés d'atténuation d'accélération d'impact des casques ont été diminuées sensiblement (de 4% à 80%). Bien que tous les casques répondent aux normes de CSA, des propriétés d'atténuation se sont avérées pour être sensiblement réduites au delà de trois impacts répétés et au-dessus de l'énergie d'impact de 40 J. En particulier, tous les casques ont montré les propriétés multiples efficaces d'atténuation d'impact à la couronne, à l'avant, et aux emplacements d'arrière; cependant, la longévité multiple faible d'atténuation d'impact était évidente sur le côté.

Acknowledgements

I would like to express my most sincere gratitude to my advisor Dr. David Pearsall, for academic supervision and professional guidance as well as his moral and financial support over the course of my studies. I also extend very special thanks to Dr. Richard Lefebvre, who so generously gave his time and whose advice has been especially beneficial towards a better thesis. I offer my greatest appreciation to Michel Desjardins, François Fecteau, and Stéphane Lamoureux at the testing laboratory of Bauer-Nike Hockey Inc. for their great technical support and creating a friendly environment for me during the six weeks data collection period: I am truly grateful.

I also would like to thank all the staff and technicians of the Department of Kinesiology and Physical Education; especially Eileen Leduc and Andrée Ippersiel for being so helpful and responsive to my needs as a graduate student.

Special thanks to my family, friends, and my fellow students from the Biomechanics Laboratory, without whose emotional support and attention I would not have been able to complete this task.

Table of Contents

ABSTRACT 2

RÉSUME 3

ACKNOWLEDGEMENTS 4

TABLE OF CONTENTS 5

TABLE OF FIGURES 7

LIST OF TABLES 8

1. INTRODUCTION 9

1.1. Nature and Scope of the Problem 10

1.2. Significance of the Problem 11

1.3. Objectives of the study 15
1.3.1. Hypotheses 15
1.3.2. Limitation 15
1.3.3. Delimitation 16
1.3.4. Dependant and Independent Variables 16

2. LITERATURE REVIEW 17

PROTECTION FROM HEAD INJURY 17

2.1. Type of Cerebral Injuries 17

2.2. Head and Brain Injury Research 21
2.2.1. Head Injury Mechanisms 24
2.2.2. Physical Factors in Head Injuries 29

2.3. Helmets 34
2.3.1. Standards for Helmet Testing 37

3. METHODS 42

3.1. Apparatus 42

3.2. Testing Protocol 43

3.3. Research Design 45

4. RESULTS 47
4.1. Multiple impact performance thresholds 47
4.2. First three repeated impacts 49
4.3. Fifty repeated impacts 57
4.4. Repeated impacts at 40 J 62
4.5. Repeated impacts at 50 J 65
4.6. Repeated impacts at 60 J 68
4.7. Peak G prediction function: regression analysis for multiple impacts 71
5. DISCUSSION 78
5.1. General Findings 78
5.2. Helmet Model 80
5.3. Site 81
5.4. Impact Energy 81
5.5. Trial 82
5.6. Liner 83
6. CONCLUSION 85
REFERENCES 86

Table of Figures

Figure 1: Acceleration-Time tolerance curve 22
Figure 2: Mechanisms of head injury 27
Figure 3: Testing apparatus 42
Figure 4: The ice hockey helmet models used in the study. 43
Figure 5: Helmet impact locations: front, rear, side, and crown 45
Figure 6: Average peak G for the helmet models at all the sites and all the energy levels. 53
Figure 7: Average peak G for the different sites for all the helmets and the energies 53
Figure 8: Average peak G for all the sites and energies for all the helmets 54
Figure 9: Comparison of peak G between 4 sites for all the helmet models 55
Figure 10: Comparison of peak G between 3 energy levels for all the helmet models 56
Figure 11: Comparison of peak G between the 3 trials, 4 sites and 3 energy levels 57
Figure 12: Average peak -G for all the trials 59
Figure 13: Average peak Gs for the 50 trials for different helmet models 60
Figure 14: Average peak Gs for the 50 trials, comparison between the sites and the trials 60
Figure 15: Comparison of peak G between the helmet models, energy levels, and the trials 61
Figure 16: Comparison of peak G between the trials and the helmet models at 40 J 63
Figure 17: Comparison of peak G between the sites and the trials at the 40 J 64
Figure 18: Comparison of peak G between the sites and helmet models at the firs 10 trials for the 40 J 65
Figure 19: Comparison of peak G between the helmet models and the trials at 50 J 67
Figure 20: Comparison of peak G between the sites and the trials at 50 J 67
Figure 21: Comparison of peak G between the sites, helmet models, and the first 5 trials at 50 J 68
Figure 22: Comparison between the sites, helmet models, and the firs 3 trials at 60 J 70
Figure 23: Average peak values for three samples of the helmet model A at 40, 50, and 60 J 72
Figure 24: Average peak values for the three samples of the helmet model B at 40, 50, and 60 J 73
Figure 25: Average peak values for the three samples of the helmet model C at 40, 50, and 60 J 74
Figure 28: A 2.5 cm gap between the front and rear padding of the helmet 79
Figure 29: Comparison of the frontal padding between two models 84

List of Tables

Table 1: Epidemiologic comparison of studies on the incidence of brain injury 14
Table 2: Diffuse brain injuries, adapted from the article by Gennarelli, 1991a. 18
Table 3: Fracture force of the skull, adapted from the article from Douglas (1993) 21
Table 4: Different Liners and padding used for the helmets 44
Table 5: The average number of impacts until peak G > 275or SI> 1500 48
Table 6: Average peak acceleration range 48
Table 12: ANOVA values for peak Gs including: helmets, sites, and the energy levels 49
Table 7: Peak G Means and Standard Deviations for Helmet Model A. 50
Table 8: Peak G Means and Standard Deviations for Helmet Model B. 50
Table 9: Peak G Means and Standard Deviations for Helmet Model C. 51
Table 10: Peak G Means and Standard Deviations for Helmet Model D. 51
Table 11: Peak G Means and Standard Deviations for Helmet Model E. 52
Table 13: ANOVA values for peak Gs including: helmets, trials, energy levels, and sites.. 58
Table 14: ANOVA values for peak Gs including all the variables at 40 J (up to 10 trials) 62
Table 15: ANOVA values for peak Gs including all the variables at 50 J (up to 5 trials) 66
Table 16: ANOVA values for peak Gs including all the variables at 60 J (up to 3 trials) 69
Table 17: Equations for the regression lines from the figures 23-27 77

1. Introduction

In sports, head injuries commonly involve impacts with other players or other equipment in the playing environment (Bishop, 1993, Pearsall et al., 2000). Injuries may result from impacts of high energy, some severe enough to fracture the cranium and / or causing concussions. With respect to the playing context of ice hockey, head injuries typically can be caused by high mass/low velocity blunt trauma (i.e. falls onto the ice or board surfaces) or low mass/high velocity impacts (i.e. projected pucks or sticks). The most frequent impact situation in ice hockey is head-to-board and body contact impacts (Bishop and Arnold, 1993).

In response to the public outcry concerning ice hockey related head injuries during the late 1960's and early 1970's, helmets were introduced to reduce the risk of traumatic head injuries. The main role of these helmets has been to act as an attenuator of linear accelerations so as to reduce the peak linear head acceleration and high-localized forces on the skull (Reid and Reid, 1981). In practice, the general construction of ice hockey helmets consists of a firm outer shell to distribute the impact force loading area and an inner liner to absorb the impact energy. Though new materials and manufacturing have evolved since then, the basic design principles of ice hockey helmets have remained unchanged since then.

To determine the effectiveness of helmet in their role, measurement of the magnitude of peak linear accelerations represents the fundamental variable of all industrial certification standards (e.g. American Standards for Testing and Materials, ASTM; Canadian Standards Association, CSA; Committee European du Normalization, CEN; International Standards Organization, ISO; Hodgson, 1991; Newman, 1993; Newman, 1983). These standards exclusive focus on qualifying the function of new helmets prior to use with respect to a minimum performance threshold; however, they do not address the issue of the viable duration of use. Recently, some standards have adopted an arbitrary five-year certification life; however, there is little research evidence indicating the functional maximal threshold for impacts or the changes in helmet properties with time and normal "wear-and-tear". To begin to answer these questions, this study will focus on the former, that is, to quantify the impact durability of helmets used in ice hockey and to define both a minimum and maximum impact attenuation profile for these helmets.

Prior to presenting the study in detail, the following text provides a summary review of the current understanding of the nature and scope of head injury forms and mechanisms, as well as, the history of head protection evaluation in sport.

1.1. Nature and Scope of the Problem

Head injuries due to accidental collisions are a leading cause of disability and morbidity in North American society (Melvin et al., 1993). It is estimated that one head injury occurs every 15 seconds, resulting in up to 2 million temporary or permanent cerebral injuries per year. About 50% of these need to be hospitalized of which 75,000-100,000 die each year. Motor vehicle accidents are the primary cause (Department of Health and Human Services, USA, 1989). Overall, head and facial injuries include 40% of all vehicle injuries (Melvin et al., 1993; Douglas, 1993).

Despite the prevalence of head injuries, the specific etiologies leading to transient or permanent cerebral damage are difficult to define (Gennarelli, 1991a; Vegso and Torg, 1991, Newman 1980). This is, in part, due to our poor understanding of the specific mechanisms causing cerebral trauma and the means of preventing cerebral dysfunction. The challenge to researchers is to better identify these mechanisms so as to inform public, government and industrial organizations on appropriate and feasible means of prevention.

From the mechanical perspective, it is self evident that high linear or angular accelerations beyond the tissue tolerances of all or part of the cranium, cerebral matters, arterial and venous (vascular-tear), glial cells, and neuron network will produce cerebral trauma (Gurdjian and Gurdjian, 1975; Gennarelli, 1991a). However, the specific acceleration threshold leading to histochemical changes is nebulous (Bishop and Arnold, 1993; Newman 1980). Our limited understanding of the mechanisms of cerebral trauma can be primarily attributed to several issues: ethical constraints on in vivo human testing; a multitude of intra-subject dependent variables (e.g. age, genetics, fitness, work environment) and the poor ability to relate cognitive dysfunctions to unspecified cerebral damage beyond general terms. Studies of cerebral trauma have predominantly been conducted by the military as well as automotive industries on cadavers, animal, and anthropometrics dummies (Gurdjian, 1975; Melvin et al., 1993; Allsop, 1993).

Gurdjian (1975) and his colleges used cadaver and animal models in their studies during 1945 to 1966. In a series of experiments a 1-kg striker was used to impact the head of anesthetized primates and record the physiological changes, impact force and head acceleration to determine the thresholds for concussion and coma. From these tests, it was observed that the skull is not rigid, but deforms with impact. Furthermore, impacts were observed to create focal or diffuse brain injuries such as tearing and hemorrhage of the brain tissue, brain swelling, and concussions. Further studies by Gurdjian (1975) and Gurdjian et al. (1962 and 1965) illustrated that skull deformations or changes in head acceleration/deceleration due to impact lead to increased intracranial pressures that resolve as shear stresses at the brain stem region above the foramen magnum. As a result, shear stresses have been implicated as the main factor responsible for cerebrovascular trauma. In addition to linear acceleration, cerebrovascular damages could also be induced by rotational acceleration (Baroff, 1998; Tysvaer and Storli, 1981; Bishop and Arnold, 1993). The threshold of human tolerance is thought to be in the order of 80-100 g peak linear acceleration (where 1 g = 1 gravitational unit $\approx$ 9.8 $m*s^{-2}$) or 1800-3500 $rad*sec^{-2}$ (300-900 $deg*sec^{-2}$) rotational acceleration (Smodlaka, 1984; Northrip and McKinney, 1988; Therrien, 1984; Bishop, 1976).

1.2. Significance of the Problem

Though not as substantial as motor vehicle accidents, it has been estimated that some 300,000 head injuries per year in the U.S. result from play in various contact sports (Smodlaka, 1984). Several surveys from various sport venues in the past three decades have raised public concern to the prevalence of head injuries, particularly concussions. Most notable, both football and ice hockey have high rates of concussion in comparison to other sports (Bishop, 1993; Rutherford and Miles, 1981). The American Football Coaches Association (1979) reported 2.6 deaths per 100,000 players, with head and neck injuries accounting for 90% of these deaths (Reid and Reid 1981). The National Athletic Injury/Illness Reporting System (NAIRS) in the incidence of concussion report (1975-1977) reported that the highest rate of concussion was in football during fall seasons

(6.1%), followed by ice hockey with a 3.7% of concussion rate (Clarke, 1991). The NCAA has stated "The fact that the most frequent injuries of the pre-1975 era (injuries to the head and face) have been virtually eliminated, attests to the effectiveness of protective equipment."; however, more alarming is the increasing rate of diagnosed concussions in players wearing full head gear (Azuelos et al. in press).

Due in large part to the physical contact inherent to the game, ice hockey players experience a high rate of head injuries. Several epidemiologic studies have attempted to identify more precisely the contributing factors resulting in head injuries. For instance, Hayes (1972) reviewed injury reports from 560 team games during 1970-71 seasons. He estimated that 61% of the injuries were suffered by forward players. Almost 70% of the injuries were attributed to body checking and stick contact. Head and face injuries (including ten concussions) accounted for 45% of total (Hayes 1972). With respect to the prevalence of concussions, a review of amateur competitive ice hockey teams from 1986 to 1990, estimated that the average rate of concussion was 4.2% of the total number of injuries (Dick, 1993) with injury rates higher during in game play (34%) than during practice (66%). Current trends can better be understood from Honey's retrospective study evaluating the incidence of brain injury in ice hockey from 1966 to 1997 has been reviewed by Honey (1998). He concluded that concussions were the most common brain injury in ice hockey. Specifically, the incidence of brain injury was found to increase with the level and hours of play. Concussion rates also varied with the level of play: in children aged 5 to 15 years the concussion rate ranged from 0.0 to 2.8 per 1000 hours of play (P1000H); for high school players it ranged from 0.0 to 2.7 P1000H; for university player it ranged from 0.2 to 4.2 P1000H; and, for elite amateurs it was estimated at 6.6 P1000H. For elite players concussion rates in real game situation is higher than during practice (table 1).

In general, concussion rates have been substantially reduced since 1966, primarily credited to the use of helmets. For instance, studies on children have shown concussion rates have decreased from 2.8, 1.5, and 0.5 in the years 1974, 1985, and 1990 respectively. However, though traumatic head injuries appear to have decreased since mandatory helmet requirements were implemented since the 1970's, the number of the players reported to suffer from concussions is still of concern. Azuelos et al. (in press) reported that prior to the 1992-93 seasons, there had never been more than 3 head injuries recorded in a single season; however, the incidence of these injuries has been rising. On average, 3.3 concussions were reported every season, accounting for 6.6% of injuries. Prior to the 1992-93 season, concussions were a rare injury; the number has since have increased, reaching eight concussions for two consecutive years (1995-96 and 1996-97) and averaging 5.0 per year (between 1992-93 and 2000-01).

Table 1: Epidemiologic comparison of studies on the incidence of brain injury (adapted from article by Honey 1998)

Reference	Cohort	Study Design	Incidence of concussion (per 1000 player-hours)	Comments
Retrospective				
Muller and Biener (1973)	all ages in Switzerland	five-years retrospective descriptive review of insurance claims	6% of injuries	no number of players at risk given
Kraus et al (1992)	intramural university hockey games n= 627 vs 648*	retrospective chart review of 2 years, 1969 (with helmet) vs 1968 (without)	2.1 (with) ŧ 4.2 (without)	"head injuries" not defined, type I error set at 0.1
Jørgensen and Schmidt-Olsen (1986)	14 Danish elite team age 16-34, n= 210	two-year retrospective, 1980s, descriptive questionnaires	6.6 in games 0.17 in practice	Bias introduced through retrospective reporting
Roy et al. (1989)	ages 12-13, n= 784*	retrospective, 1985-86 questionnaire, phone and some observation	1.5*	1 concussion in league with checking 1 in league without
Pförringer and Smasal (1987)	West German Bundesliga 1, n=88	retrospective, 1986-87 interview	17	denominator for risk of injury unavailable
Prospective				
Daffner (1977)	n=250, ages 5-17 n= 75, ages > 17 n= 40, semipro	prospective, 1974-76 report by team physician on hours at risk, no	0ŧ	small cohort, no data serious injury
Sutherland (1976)	n= 707, ages 5-14 n= 207, high school n= 25, university n= ?, one pro team	prospective, 1974-75	2.8; 2.7; (3% of injuries); (5% of injuries)	denominator for risk of injury not vailable for university or pro
Pelletier et al. (1993)	Intercollegiate, men ages, 19-25, 9424 player-games	prospective, 1979-85 review of injury report	1.5*	games only
Lorentzon (1988)	elite Swedish team	prospective, 1980- 85 physician-observed	0.2 include practice hours	rate could be as high as 4.4 if all occurred at game
Lorentzon (1988)	Swedish national team	prospective, 1984-85	0	only 240 player-games
Brust (1992)	ages 9-15, n=150	prospective , 1990-91	0.5ŧ	"blow to head" not defined
Stuart et al. (1995)	junior A, age 17-20, n=25	prospective, 1990-93	0	---
Stuart et al. (1995)	ages 9-14, n= 66	prospective, 1993-94	0	small cohort
LaPrade et al. (1995)	NCAA division I team total practice-hours 18,584 total game-hours 1,008	4 years prospective as review by trainer, 1990s	0.2 include practice hours	rate could be as high 3 if all occurred at game

*calculated data, ŧ = articles in which "concussion" is not mentioned.

As a consequence, the ability to assess the functional capacity of helmets to protect the brain from injury is necessary if one wish to speak to the health concerns rose. Research in sport protective headwear endeavors to replicate the dynamic mechanisms that result in cerebral trauma, particularly with respect to concussions. To date, numerous techniques and standard testing protocol have been created. Principally, all these approaches possess defined impact environments where predetermined energies and helmet orientations are controlled. Stemming from these collisions, various dynamics measures from these impacts are evaluated and compared to established threshold tolerances. Most helmets are evaluated immediately post-manufacturing to ensure product quality and fulfillment of impact standards for certification. Unknown, is the capacity of current helmets with repeated use and ageing to retain their functional ability to reduce head accelerations within safe limits.

1.3. Objectives of the study

Intend of this study was to evaluate the long-term mechanical effectiveness of ice hockey helmets sustaining focal impacts. This study focused on quantifying the durability of protective helmets designed to reduce the severity of injuries such as fractures and concussions in sport and recreational activities. Specifically, this research addressed the effectiveness of ice hockey helmets after multiple impacts at defined energy levels. This also addressed the question of how many impacts a helmet can tolerate without changing its mechanical behavior. Thus, using "aged" helmets could increase the injury potential in the head. This information can be a valuable guide to standards defining the use-life of helmet products.

1.3.1. Hypotheses

It was hypothesized that:

1. The ice hockey helmets impact attenuation ability will decrease significantly with increased number of multiple impacts per site;
2. The ice hockey helmets impact attenuation ability will decrease significantly with increased impact energy;
3. The ice hockey helmets impact attenuation ability will vary significantly between impact sites with regards to increased number of impacts and increased impact energy;
4. The impact attenuation ability will vary significantly between ice hockey helmet models.

1.3.2. Limitation

The limitation of the study will be:

1. The tests will not perform on human subjects.
2. All tests will be performed on a surrogate anthropometrical form by using a drop apparatus.
3. Only five models of hockey helmets will be tested.

1.3.3. Delimitation

The delimitations of the study were:

1. All tests were conducted in the Product Testing Laboratory of Bauer-Nike Hockey Inc. (Saint-Jerome, Quebec, Canada).
2. A monorail drop testing apparatus was used.
3. For all tests, the condition of the test was determined using the CSA Standards CAN/CSA-Z262.1-M90.
4. All tests were conducted at room temperature.
5. The helmet models were the latest product models.
6. The helmet models were all size "Large"
7. The helmet models were chosen from the most accessible brands.
8. Only blunt impacts were tested.

1.3.4. Dependant and Independent Variables

The main dependant variables measured in this study were: impact time, the Severity Index (SI), and the peak linear acceleration (peak G) of the headform. The independent variables were drop velocity, impact height, impact location, kinetic energy, helmet model, helmet samples and repetitive measurements.

2. Literature Review

Protection from Head Injury

To better appreciate the topic of protection form head injury, the following section will provide relevant background information. This will include discussion of the types of cerebral injuries (focusing on the various clinical presentations of concussions), as well as a brief synopsis of the history of research pertaining to head injuries (with an emphasis on the mechanisms of head injury and related physical factors) and the evaluation of head protective equipment as used in sport.

2.1. Type of Cerebral Injuries

Injury of the cerebrum has several forms ranging from mild concussions to diffuse white matter injuries. The severity and duration of symptoms often correspond to the extent of neurophysiologic disturbance. Cerebral disruption may result directly from insult to cerebral tissue or secondary to other cranial complications such as vascular lesions and skull fractures.

The most common type of head injury is the mild concussion, often completely reversible. Symptoms include confusion, disorientation, and possibly a brief duration of post-traumatic and retrograde amnesia but no unconsciousness. The classical cerebral concussion includes loss of consciousness immediately after an impact. Consciousness should be regained within 24 hours, symptoms are reversible and amnesia is typically present. The duration of amnesia indicates the severity of the concussion. Gennarelli (1991[b]) reported that no lesion occurred in 36% of these injuries. He observed that the damages distributed by regions included: 10% of cortical contusions, 10% vault fracture, 7% basilar fracture, 35% depressed fractures, and 36% multiple lesions. At the end of the first month, 95% of patients had full recovery, 2% of them may have moderate and 2% retain severe cognitive deficit (Melvin et al., 1993).

Another category of cerebral injuries includes *diffuse brain injury*. The symptoms of this type of injury include: amnesia lasting for days, mild to moderate memory deficit, mild motor deficit, and may be accompanied with de-cerebrate posture. At the end of first month, 21% of patients have good recovery, 50% have moderate or severe deficit, 21% have vegetative survival, and 7% are fatal (Melvin et al., 1993). The most severe form of this is termed *diffuse axonal injuries* (DAI). DAI occur when a large amount of axons in the brain hemispheres and sub-cortical white matter are physically disrupted. Abnormal brainstem symptoms indicate that axonal disruption can be extended to the distal of midbrain and brainstem (Gennarelli, 1991b). De-cerebrated posture, loss of consciousness for days or weeks, severe memory and motor deficits, and post-traumatic amnesia lasting for days are the signs of DAI. Microscopic studies indicate that axonal tearing of white matter occurs in the hemispheres and degeneration of long white matter projector tracts down into the brainstem. Small hemorrhages at the corpus callosum, superior cerebellar peduncle, or periventricular region related to this condition may be detected by high resolution CT scans. At the end of the first month, 55% of patients would have died, 3% may have vegetative life, and 9% would have severe deficits (Gennarelli, 1991b). Table 2 shows the signs of various diffuse brain injuries.

Table 2: Diffuse brain injuries, adapted from the article by Gennarelli, 1991a.

			Diffuse Axonal Injury		
	Mild Concussion	Cerebral Concussion	Mild	Moderate	Severe
Loss of Concussion	None	Immediate	Immediate	Immediate	Immediate
Duration of Coma	None	< 6 hours	6-24 hrs	Days	Days-Weeks
Decerebrate Posturing	None	None	None	Rare	Common
Posttraumatic Amnesia	Minutes	Min/hr	hr	Days	weeks
Memory Deficits	None	Min	Mild	Moderate	Severe
Motor Deficits	None	None	None-Mild	Mild-Moderate	Moderate-Severe
Outcome at 3 Months			%		
Good Recovery	100	95	63	38	15
Moderate Deficit	0	2	15	21	13
Severely Disable	0	2	2	12	14
Vegetative	0	0	1	5	7
Dead	0	0	15	24	51

In contrast to cerebral disturbance resulting from the impact sustained directly to cerebral tissues, cerebral disruptions can be also induced indirectly such as by increased intra-cranial pressure due to brain swelling commonly resulting from increased intra-vascular blood within brain (Gennarelli, 1991b), or cerebral edema due to an overall increase in tissue fluid. About 4% to 16% of all head-injured patients and 28% of head-injured children have brain swelling, but mortality rate in adults is 33-50% and in children is 6%. Examples of the latter include *acute subdural hematoma* (ASDH), *extradural (epidural) hematoma* (EDH), and *intracerebral hematoma* (ICH). ASDH result from laceration of cortical blood vessels, large contusion bleeding into the subdural space, or tearing of the veins that connect the subdural space to the dural sinuses. ASDH occurs in 30% of patients with severe head injury with 60% mortality rate. The ASDH are more frequently observed in athletes than EDH because most athletic head injuries are from lower inertial loading (Bruno, 1991). The mortality rate in pure ASDH is around 20% and if associated with other brain injuries is around 50% (Bruno, 1991). Patients with simple ASDH usually retain consciousness. CT scan imaging is the best way to diagnose the location and the size of hematomas (Melvin et al 1993, Gennarelli, 1991b). Alternatively, *extradural hematoma* (EDH) result from entrapment of a meningeal artery in its bony groove with or without skull fracture. Loss of consciousness at the time of injury is a classic sign with variable recovery time. The mortality rate ranges vary from 15% to 43%. Brain decompression by surgical removing of the clot is necessary (Melvin et al., 1993). *Intracerebral hematomas* (ICH) are usually caused by sudden accelerations/decelerations of the head. The ICH is a homogeneous pooling of blood within the brain tissues and can be diagnosed by CT scan. ICH and contusions can be the result of combined impacts and impulses. Penetration impacts and depressed fractures can cause ICH as well. Post-traumatic amnesia, confusion, and persistent headache are signs of ICH. Hemorrhages start superficially and can extend deeply into different parts of the brain without causing unconsciousness in the patient. The incidence rate is from 4% to 23% and the mortality rate is from 6-72% depending on the span of the injury (Melvin et al 1993, Gennarelli 1991b).

Often associated with the above injuries are *cerebral contusions*. These are classified into two categories: coup contusions, occurring at the impact site, and countercoup contusions that occur at the opposite side of the cerebrum from the impact site. Cerebral contusions are usually associated with other lesions and skull fractures in 60-80% of cases. Hemorrhage, edema, infarction, and necrotic areas in the brain can also be observed. Mortality rate ranges vary from 25% to 60%. In adults over 50 years and in the patients suffering a coma, the mortality rate is even higher (Melvin et al., 1993).

It should be noted that cerebral pathologies are often but not necessarily corresponded to fractures of the skull. In mechanical terms, *skull fractures* are caused by focal contacts of 2-7 ms duration or with onset of from 20,000 to 50,000 G/s (Hodgson and Thomas, 1972). The type and extension of the fracture depends on the material properties of the skull, the magnitude and direction of impact, the size of impacted area, the area, the shape and the mass of impacter, and the thickness of the skull bone. However, some research has suggested that mineral content and pulse duration do not have significant affect on fracture force (Allsop, 1993). Normally, fractures occur along the same direction as the impacted force. They occur readily with tensile stress than with compression stress. Fractures begin at the region of induced tensile stress. The correlation between simple linear fractures and neural injuries is not consistent. The fracture itself does not have clinical importance, except for basal and depressed fractures, which may indicate damage to the meningeal vessels, dura, or cerebral tissue resulting in brain contusion. The incidence rate for depressed fracture is 20 per 1,000,000 per year and the mortality rate is 11% because of the central nervous system injury (Gennarelli, 1991a; Allsop, 1993; Melvin et al., 1993). The incidence rate for basal fracture varies from 3.5% to 24% due to the difficulties involved in diagnosing such a fracture with a clinical examination. Hodgson and Thomas (1972) showed that up to 80% of those having skull fracture, have also concussion. Table 3 shows the range of force that induce skull fractures.

Table 3: Fracture force of the skull, adapted from the article from Douglas (1993)

Bone	Force			Impactor area (cm^2)	Author (reference)
	Range (N)	Mean	Sample size		
Frontal	2,760-8,850	4,930	18	6.45	Nahum (1968)
Frontal	4,140-9,880	5,700	13	6.45	Schneider (1972)
Frontal	2,200-8,600	4,780	13	20-mm dia bar	Allsop (1988)
Frontal	5,920-7,340	6,370	4	6.4-mm dia bar	Hodgson (1973)
Frontal	8,760-8,990	8,880	2	25.4-mm dia bar (sagittal)*	Hodgson (1973)
Frontal	N/A	6,550	1	50.8 mm dia bar (90 durometer, sagittal)*	Hodgson (1973)
Frontal	N/A	6,810	1	203 mm radius hemisphere	Hodgson (1971)
Frontal	4,310-5,070	4,690	2	76 mm radius hemisphere	Hodgson (1971)
Frontal	N/A	5,120	1	50.4 mm dia bar (sagittal)*	Hodgson (1971)
Left frontal bones	2,670-4,450	3,560	2	25.4 mm dia bar	Hodgson (1971)
Temporoparietal	2,215-5,930	3,490	18	6.45	Nahum (1968)
Temporoparietal	2,110- 5,200	3,630	14	6.45	Schneider (1972)
Temporoparietal	2,500-10,000	5,200	20	5.07	Allsop (1991)
Parietal	5,800-17,000	12,500	11	50	Allsop (1991)
Zygomatic arch	930-1,930	1,450	11	6.45	Schneider (1972)

* major axis of bar parallel to sagittal plane.

2.2. Head and Brain Injury Research

The primary objective for experimental research on head and brain injuries has been to reproduce the anatomical, physiological, and functional responses as those described clinically. This information was needed to improve our understanding of injury mechanisms, means of prevention as well as assist in the design of protective equipment.

The early research to identify human head tolerance with respect to impacts was started by German physicians in 1914 (Reid and Reid, 1981). The first non-penetrating head impact experiments were conducted in primates and then non-primate species by Denny Brown and Russell in 1941 and 1945. Using unconstrained head impacts to permit head motion, researchers observed that concussions occurred less than if restrained. Starting in 1944, Gurdjian and colleagues used various strikers with different mass, contact area, and velocities to collide with the head of cadavers and anesthetized primates so as to determine the thresholds of head injuries. Also, several predetermined impact locations on the heads were investigated. By using different transducers and high-speed camera they recorded head acceleration, impact force, head motion, changes in intracranial pressure, and physiological changes certain impulse patterns were identified as being particularly hazardous; such as, long acceleration phases with abrupt deceleration (i.e. jerk). The produced injuries in the animal-model and cadaver based studies were similar with those observed in clinical studies (Melvin et al., 1993).

These initial studies consistently demonstrated that impacts to the unrestrained head produce both rotational and translational acceleration. In addition to the local deformation of the skull due to direct impact, the resulting accelerations in time experienced by the head have been identified as primary injury factors (Newman, 1983). This was best demonstrated by Lissner (1960) who initiated the development of the Wayne State Tolerance Curve (WSTC) (fig 3). The WSTC is based on head translational acceleration mainly in frontal plane, which is thought to cause shear strain injury in the brainstem region. The relationship between acceleration and pulse duration indicated that the tolerance level decrease as time duration increases. Initial, the curve was based on only six samples and durations of 1-6 ms. Subsequently, this was supplemented by Gurdjian, Lissner, and Patrick (1962) in a series of experiments on animals, human cadavers, and volunteers to find thresholds head acceleration for linear skull fractures and resultant concussions. Injury thresholds were found to be on average approximately 160 G for human linear skull fracture and 90 G for concussions. 80 G was identified as the injury threshold for long impulse durations based on rocket sled tests with human volunteers in the military. Head accelerations of 80G or less caused little or no observable injuries, though, occasionally tearing of vascular vessels producing intracranial hemorrhages could still exist (Gurdjian, 1975; Melvin et al., 1993).

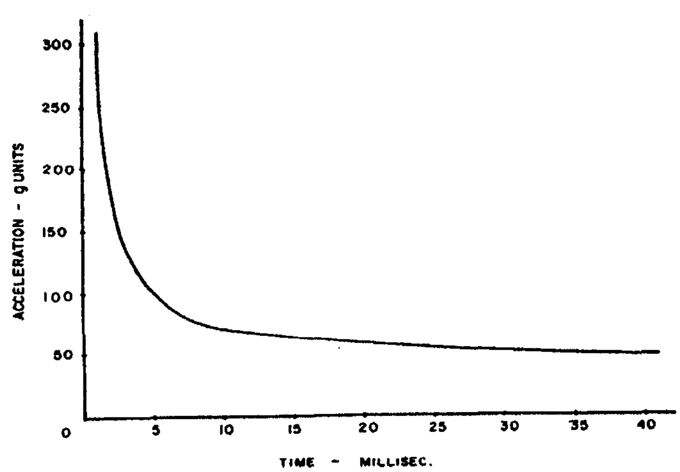


Figure 1: Acceleration-Time tolerance curve (Gurdjian, 1975)

One of the criticisms of the WSTC was that "a single point on an applied pulse cannot accurately define the response of a physical structure to that pulse" (Gadd, 1966). To better estimate the dynamic relationship with injury, the Gadd Severity Index (GSI) was proposed as a "weighted impulse criterion". The GSI calculates the cumulative impulse according to:

$$GSI= \int a^{2.5} (t)dt \qquad \text{(Equation 1)}$$

where

a = instantaneous acceleration center of gravity of the head

t = impact pulse duration.

The threshold GSI value to produce head injury was defined as 1000. For example, 1000 GSI is equivalent to an acceleration magnitude of 100g for 10ms duration. The principal advantage of the GSI was that it eliminated differences in judgment and permitted repeatable and comparable test result in different laboratories. The other advantage was using the power of 2.5 for acceleration that solves the dilemma of having sharp pulse (i.e. triangular pulse) versus flat pulse (i.e. square pulse), but with same G–time area and different injury hazards (Gadd, 1966). Subsequently, the GSI was modified so that the "conventional average waveform level for acceleration" was replaced by the "effective acceleration". Termed the Head Injury Criteria (HIC), this modified calculation integrates the total duration of the acceleration pulse (Newman, 1980). HIC is calculated as:

$$HIC= [\int a\, dt / (t_2-t_1)]^{2.5} (t_2-t_1) \qquad \text{(Equation 2)}$$

where t_1 and t_2 are arbitrary onset and offset of pulse to maximizing the HIC.

Injury threshold values for serious to fatal injuries of 1000 were initially proposed. Schneider and Zernicke (1989) concurred with these HIC values from mathematical modeling. However, cadaver studies using ink perfusion in the brain suggested the injury threshold value of 1500 (Mellander, 1986). This criterion was developed for impact loading rather than impulse loading. Another variation, termed the Japan Head Tolerance Curve (JHTC), was introduced by Ono in1980. JHTC has slightly longer impact duration (i.e. more than 10ms) than the WSCT. The JHTC threshold values to cause skull fractures are slightly higher than the value for concussion (Melvin et al.,1993; Gurdjian, 1975; Newman, 1983; Advani et al., 1982; Mellander, 1986).

Though high translation accelerations have been clearly shown as a causal mechanism for concussions (Newman, 1980); several authors have also stated that high levels of rotational acceleration may be equally or more hazardous (Gurdjian, 1975; Gennarelli, 1991b; Mills and Gilchrist, 1997). For instance, cerebral injuries could be induced by 60° of angular displacement within 11-22 ms, and peak angular accelerations up to 2×10^5 rad/s^2 without associated skull fractures (Gennarelli et al., 1982). It was postulated that rotational accelerations (due to direct or indirect impacts} produce relative motion of the brain with respect to the skull resolving as shear stresses in turn creating lesions within and around the brain tissue (Gurdjian, 1975). Supporting this notion, Ommaya and Hirsch (1971) reported that rotational acceleration of the head greater than 1800 rad/s^2 corresponds to a 50% possibility of concussive head injury.

2.2.1. Head Injury Mechanisms

The research findings noted above present the general mechanisms responsible for head and brain injuries. In the following, a more comprehensive analysis of the specific dynamic mechanisms and tissue interactions of the skull and the brain will be discussed.

By definition, an impact is a dynamic event where upon momentum is transferred from the collision of two or more bodies. The magnitude and changes in energies are related to the nature of the motion in time (i.e. trajectories, velocities, accelerations). The developed energy can cause deformation, rotation, and acceleration of the head. For deformable bodies, the magnitude of the change in momentum (i.e. impulse) is proportional to the resultant forces and the duration of contact. This impulse creates stresses (i.e. compression, tensile, and shear) within the tissues leading to tissue deformations or strains in numerous directions relative to the body's coordinate system (Bandak, 1995). Given the extreme difference in stiffness properties between the skull and the brain, the dynamics of energy transfer are complex and difficult to predict. Consequently, the viscoelastic brain can be made to move relative to the skull. This differential movement creates high tissue interface stresses that may create focal or diffuse brain injuries such as tearing and hemorrhage of the brain tissue and cerebrovasculature, brain swelling, and concussion (Baroff, 1998; Gurdjian and Gurdjian, 1975). These hazardous conditions can also be generated by indirect impacts (i.e. impacts to the body and not the head). For instance, an impact to the upper trunk can generate high head acceleration due to the inertial lag of head (Bandak, 1995).

A resultant head injury depends on the interaction of several factors including: (1) the point of application of force, (2) the nature of force distribution, (3) the head motion as a result of the force acting on it, (4) the relative motion of the brain or part of it inside the skull, and, (5) the local stretching of the brain stem and spinal cord because of the motion produced in head and neck junction (Melvin et al., 1993). The magnitude of applied forces is a very important factor to make each of these mechanisms worst. Injury mechanism paradigms describing these factors have been proposed (Fig 2, Gennarelli, 1991a). This includes differentiating between the rates of mechanical input on a continuum from slow (static) to fast (dynamic). Static loading (i.e. 200 to 500 ms or longer time duration) is very uncommon in sport injuries unlike dynamic loading (i.e. less than 200 ms in duration). Dynamic loading may be classified in two forms: (1) impact (contact loading), and (2) impulse (inertial loading) (Gennarelli, 1991a).

Contact loadings are often oblique and result primarily in skull fractures. Contact loadings also produce rapid changes in intercranial volume that in turn rapidly transmit cerebral tissue stresses alike to shock waves at the speed of sound. Several variables influence the location, type and extent of fracture such as the magnitude and direction of force, contact area, thickness of skull, and material property of the skull.

If the magnitude of the force is sufficiently large, then the skull may bend inwards at the site of impact. The extent of skull deflection depends in part on the local thickness of the bone; for instance, thickness can vary from 2.2-6.0 mm at the temporal (squamous) region to 5.0-9.6 mm at the skull vertex (parietal bone) in the adult male. Hence, the skull's bone strength is not uniform i.e. thinner bone regions will possess lower ultimate strength properties. Several studies have been conducted to determine the human skull fracture forces and skull stiffness. Allsop et al. (1988) found the mean stiffness of human skull by impacting cadavers' head. The stiffness magnitudes in the frontal, temporoparietal, and parietal area were 1000, 1800, and 4200 N/mm, respectively, using impactors with the areas of were 20mm diameter, 6.45 cm^2 (25.4mm diameter), and 50 cm^2. In addition, the failure strength of skull bone has been documented to range from 4.5-11.9 kN in dynamic conditions, and from 8.8-14.1 kN in quasi-static conditions (Yoganandan, 1995). Given the inhomogeneous bone strength distribution of the skull, fractures do not necessarily occur at the point of impact (Gennarelli, 1991a). For instance, a supraorbital impact site may result in a fracture at the base of the skull where as an upper frontal impact may produce a fracture to the cranial vault.

Nonetheless, the location of a fracture provides important evidence of the impact direction. Yoganandan et al. (1995) demonstrated the fracture behavior of a skull by submitting cadaver heads to quasi-static and dynamic loading. They observed that fracture widths were smaller on the impact side in comparison to more remote skull locations. The deflection range for quasi-static loading was 7.8-16.6 mm and for dynamic loading was 3.44-9.74 mm. For instance, for the case #7, which was impacted at vertex, the fracture occurred at frontal side rather than vertex (5.77 mm wide). However, there is no report on the wide of fracture at each side, only the average deflection is reported. Their finding was similar to that of Gurdjian (1975).

Research has repeatedly shown that concussions can be produced without associated skull fractures. Nonetheless, the reverse is not necessarily the case i.e. skull fractures are often implicated with cerebral injury.

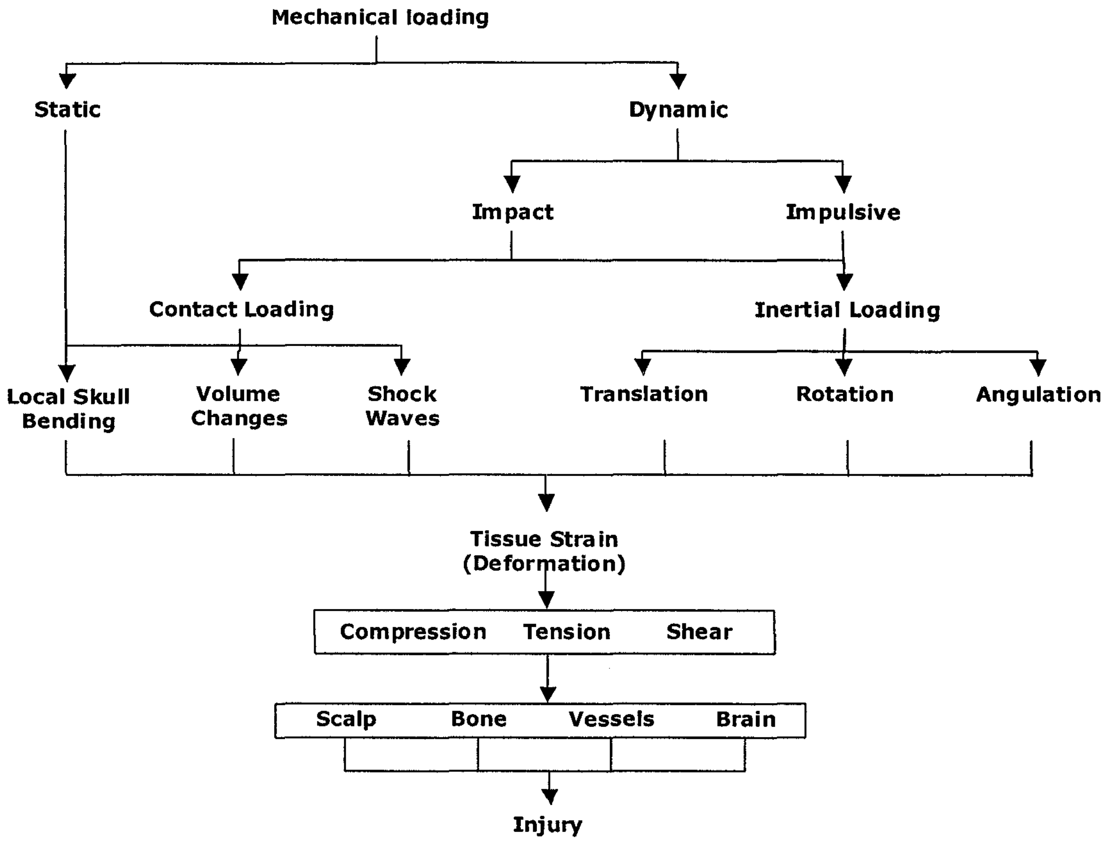


Figure 2: Mechanisms of head injury (adapted from Gennarelli 1991a)

Contact loading can also cause volume changes without causing a fracture. This occurs due to mass motions or relative movement of the brain and brain stem. A rigid skull in contact with relatively movable brain can cause pressure gradients. On the impact side of the skull, high pressure due to inward bending of the bone, and at the opposite side low pressure will be produced. First the brain will move to the impact side because of inertia, causing coup contusion, and then it will rebound in the direction of the force, causing countercoup contusion due to impact with the rough and irregular bone on the interior side. Moreover, the existence of the foramen magnum induces upward/downward movement of the brainstem and posterior fosa structure upon impact. Concentrated stress in the foramen magnum can be transmitted to the spinal column causing spinal cord injury (Gurdjian and Gurdjian, 1975). Nusholtz et al. (1995) explained this phenomenon using a simple model. They found that in pre-stressed head due to accelerations, impact would cause cavitation. The cavitation can be on the impact side, opposite the impact side or anywhere else in the brain. The cavitation also depends on cerebrospinal fluid (CSF). In a vertical impact, the CSF will outflow to the spinal column thereby reducing brain pressure and consequently, less injury to the brain; however, an elongation distortion downward through the foramen will occur (Johnson, 1991) and the crowding of the brain toward the foramen magnum may cause contusions of the brain stem (Gurdjian, 1975).

The second type of dynamic loading, the inertial loading, caused by an impulse, can lead to acceleration or deceleration of the head. Acceleration occurs when the velocity of the body increases due to impact with a faster moving body, but deceleration occurs by sudden stopping or slowing down enforced by a stationary or slow moving body (Gurdjian, 1975). They both induce the same physical input (except they are in the opposite directions) and they are both accompanied by compression. However, the effects may differ. Internal cerebral compression is minimized when the head can move freely. In the case of a deceleration impact, generally, there is a greater amount of coup lesion and compression than during an acceleration impact. Deceleration can cause extensive deformation and compression as the head stops. Because due to inertia, the brain continues to motion and the stationary skull stops it and consequently, the kinetic energy of the brain produce contusion. Fractures may occur in deceleration impacts. Impacting an immobile head can easily produce a coup contusion and if the head is stationary on a solid hard surface, a countercoup contusion may occur at the opposite side if no rebound occurs (Gurdjian and Gurdjian, 1975).

Acceleration during impact is a biphasic motion, first a long ramp like acceleration phase followed by an abrupt deceleration phase (Melvin et al., 1993). Acceleration impacts create an initial compression as the body is made to move due to inertia but with more acceleration, compression is insignificant (Gurdjian, 1975). Then the head is accelerated in the direction of the impact force, obviously, counter coup lesion may be significant. When head is accelerated with impact, there is no time for fractures to occur. Inertial loading due to relative movement of the brain within the skull can induce shear and tensile forces in the brain and consequently, can cause various types of injuries, such as: large subdural hematomas due to subdural bridging veins torn, shearing damage to white matter axons, contusion of the brain surface in contact with the skull during relative rotation of the brain, bleeding, increase in intracranial pressure, concussions, or any other type of lesions may occur (Gennarelli, 1991b). Each of these injuries may cause a series of secondary events; for instance, a hematoma can cause a local mass effect and increase in local tissue pressure and pressure gradient within the brain. The pressure gradient may shift the brain and/or herniate it toward the foramen magnum. This herniation may cause compression of the brain stem and damage the vital controlling center of respiratory and cardiac. Therefore, not only the primary also the secondary mechanism of the injury is important (Gennarelli, 1991b).

The type of the acceleration/deceleration delivered to the head can have different influences in the type and extent of injury (Gennarelli, 1991a). Acceleration/deceleration can be rotational, translational, or a combination of both. The study by Gennarelli (1991b) showed that in maximum translational acceleration it was impossible to induce concussion but it was easy in rotational acceleration. He concluded that focal lesions can be caused by translational and diffuse injuries occur by angular Acc/Dec. Later study by Gilchrist and Mills (1996) showed that angular acceleration is known to be the leading cause of diffuse axonal injuries. Schneider and Zernicke (1989) also reported that the risk of a head injury, with an angular acceleration is greater than with linear acceleration. Researchers show that at a very high level of inertial loading, such as head-on collision (hyperflexion injuries), subdural bridging veins can be ruptured and large size hematomas may occur. These changes can cause death or very serious diffuse injuries.

2.2.2. Physical Factors in Head Injuries

Head injuries are generally classified into two groups: diffuse injuries and focal injuries (Gennarelli, 1991b). Diffuse axonal injuries are attributed to high-mass/low-velocity blunt traumas, which create head and brain accelerations, resulting in concussions and brain swelling. Focal injuries are induced by low-mass/high-velocity impacts and may cause skull fracture, arterial damage both above the dura (such as epidural hematomas) and below the dura (subdural hematomas), intracerebral hematomas, and contusions.

The head is free to move, but its movement is restricted by its attachment to the body via neck. Therefore, upon impact the head will move linearly, but its attachment will lead to rotations. Alternatively, acceleration of the body may be transferred to the head causing inertial acceleration or deceleration. Melvin et al. (1993) explained that "the impact to the unrestrained head always produces both rotational and translational acceleration." High levels of rotational acceleration may cause loss of consciousness greater than high levels of translational acceleration. Some studies have shown that rotational acceleration; in both direct and indirect impacts induce concussive brain injuries while producing shear strains, which in turn induce shearing deformations without any volume change in brain tissue. Gurdjian found that head rotations might cause relative motion of the brain with respect to the skull and result in injury. Ommaya in 1985 showed that the magnitude of rotational accelerations necessary to induce concussive head injuries in direct impacts is approximately half the magnitude of that needed in an indirect impact (Melvin et al., 1993). Ommaya and Hirsch in 1971 reported that if rotational acceleration of the head is more than 1800 rad/s^2, there is a 50% possibility of concussive head injury. Ommaya and Gennarelli in 1974 reported that diffuse brain injuries are related more to rotational accelerations than pure translational accelerations. However, Ono et al. (1980) reported that pure translational acceleration is capable of producing a concussion. The above authors agree that the resulting head injury will depend on the variation of acceleration over time (Newman, 1983). Therefore, the combination of rotational and translational accelerations and local deformation of the skull are implicated as injury mechanisms (Melvin et al., 1993).

Under a focal impact, the applied force can accelerate or decelerate the head causing injury; both of these conditions may result in compression. However, in a deceleration impact, compression will be greater due to a great amount of deformation. During impact acceleration, the inertia of the head will induce some initial compression. Compression itself also causes acceleration and deceleration in the brain.

Ommaya in 1985 classified the impact injuries in five categories: (1) Direct (coup) contusion of the brain from skull deformation, fracture, or movement of brain against rough interior surface of skull; (2) Reduced blood flow due to infarction or pressure; (3) Indirect (countercoup) contusion at the opposite site of the impact; (4) Relative motion of the brain hemispheres and the skull that induces tissue stresses; or (5) Subdural hematoma because of ruptured bridging vessels between the brain and the dura matter. The last three categories maybe induce by direct impact or high inertial accelerations (Melvin et al., 1993).

Both focal impacts and whiplash (impulsive) impacts induce deformation and stresses in the brain and its structures. Gurdjian (1975) reported that the brain is relatively incompressible and it appears more solid than viscous. The brain is a viscoelastic mass surrounded by CSF (Gurdjian and Gurdjian, 1975). Its viscoelastic components react differently to the applied stresses. For instance, the dura matter folds and protrudes inward to make falx cerebri, falx cerebelli, and tentorium cerebelli; these structures help to limit excessive brain rotation. The CSF in subarachnoid layer works as a shock absorber. This will affect the relative motion of the brain and skull. During impact the CSF will induce pressure gradients. It has been proposed that the homogenous pre-capillary system of the brain is equipped with a load bearing property that effectively remove the loads away from the brain's neural tissue (Bandak, 1995).

When the head is loaded, it will experience changes in each part of its structure. These changes can be one or combination of two or more parameters: (1) compression, tension, and shear of scalp, (2) elastic deformation of the skull, (3) mass motion, and (4) relative motion of the brain with distribution of inertial stress and inducing of negative and positive pressure inside the skull (Gurdjian and Gurdjian, 1975).

The scalp has a high tensile property that protects the head during low velocity/low energy impacts. Injury of the scalp will depend on the shape of the impacter and the magnitude of applied stress. It absorbs some of the impact energy of focal impact.

Elastic deformation of the skull occurs with focal impact. If the applied stress is under the failure strength of bone but high enough to deform it, the skull will be deflected inward. In the condition that the head is stationary, the brain tissue may be contused on the impact side. If a linear fracture does not occur, the injury may be extended by an induced pressure gradient. When bone is unloaded and out-bended, it can separate the dura and cause epidural hematomas. If a fracture occurs, vessels of the epidural or dural lining will tear and epidural and subdural hematomas will occur. Dura tear and mass motion of the brain may entrap the pia matter and consequently, can cause vascular tear and aneurysms. Fractures provide an entrance for foreign objects or scalp, it also cause contusion of the brain tissues under the fractured area. Entrapment of the basilar or vertebral artery can occur in existing of basal fracture due to focal impacts to the occipital area. Mass motion of the posterior part of the brain and brainstem in postero-anterior direction occur. During motion they can entrap in fracture lines and contuse.

Development of pressure gradients occurs during focal blunt impacts and/or inertial stresses. Inertial stress occurs with focal or indirect impacts. The rigidity of the skull and the movable brain inside CSF are responsible for the observed pressure changes. Upon impact a high pressure area develops on the impact side and a low pressure area at the opposite side of the impact. This occurs because of inertia of the brain resists to the impact force during the first 2 ms. This movement may cause vessels tears and some contusion around the brain. Then, the brain moves in the direction of force and may cause countercoup cavitation, which was explained earlier. This lesion is due to the impact of brain with the skull or flattening of the skull by the brain. Negative pressure is not necessary to induce cavitation. With inertial loading, both coup and countercoup lesions occur due to the slapping effect (a sharp blow) of the skull and the motion of the brain against the rough surface of the skull respectively. A high translational acceleration results in a low pressure gradient but brain contact with the skull is the main reason for countercoup lesions. In high acceleration coup lesion is not significant, since there is no enough time for the brain to move towards the impact side. However, this is significant for countercoup lesions, especially when a fracture does not occur and the CSF moves towards the impact side. In general, relative movement of the brain with respect to the skull can cause contusions, intracerebral bruises and countercoup lesions (Gurdjian and Gurdjian, 1975; Gurdjian 1975).

Rear-end, head-on, and lateral impacts (in automotive accident) are all indirect head injuries that may result in concussions, intracranial hemorrhages, and bruises of the brain surface and deep tissues due to the induced angular acceleration. Velocity differences between the head and the impacted body cause a momentum change in the head (as described before). Momentum causes deformation and internal stresses inside the skull, which can cause variety of lesions. The lesion side depends on the direction of the impact loading on the body. In head-on collisions, if the head is stopped with a solid material, coup or countercoup lesion will occur (Gurdjian, 1975). Lateral impacts; also produce pressure gradients in the brain. These gradients are proportional to the head acceleration components. The greater component is in the direction of impact and produces a positive pressure (compression) near the impact site and a negative pressure on the opposite side of impact (tension) (Nahum et al, 1981). The rate of mortality due to lateral bending is very high, second order to head-on injuries.

Concussions are one of the most well-known injuries that occur in fast-acceleration/short-duration impacts. Incidences of concussions occurring upon contact with a windshield are a small number. Occupants with shoulder belts do not strike their head; therefore, concussion will not occur (Hodgson and Thomas, 1972). On the contrary, Ommaya et al. (1964) explained that concussions occur more readily when the head is permitted to move freely on the neck. This occurs usually following occipital blows with linear accelerations of 700 g. When the neck is restricted from moving (using a collar strap), it is difficult to induce a concussion; an acceleration of 900 g would be needed (Ommaya et al., 1964).

Concussion or loss of consciousness is one of the well-known brain diffuse injuries immediately after impact (Melvin et al., 1993; Gurdjian, 1975; Gennarelli, 1991b). Concussion is found to be the prevalence of accidental head injuries in North America. Head and facial injuries make up a total of 40% of all vehicle injuries. Studies by Gennarelli showed that three in four automobile brain injuries are in the form of diffuse injury with the remaining representing by focal injuries, conversely, fall victims had the opposite distribution. He showed that subdural hematomas and diffuse axonal injuries were the most common cause of death due to brain injury (Melvin et al., 1993).

Concussions have also become a major concern especially in sport where it appears to be the most prevalent head injury. Both football and ice hockey have a high rate of concussion (Bishop, 1993). Football has the highest rate of incidences of head and neck injuries followed by baseball, which has highest incidences of head and face injuries in players from 5 to 14 years old (Rutherford and Miles, 1981). The American Football Coaches Association (1979) reported 2.6 deaths per 100,000 players, head and neck injuries accounted for 90% of these deaths (Reid and Reid 1981). The National Athletic Injury/Illness Reporting System (NAIRS) in the incidence of concussion report (1975-1977) reported that the highest rate of concussion was in football during fall seasons (6.1%) and hockey was in the second place with 3.7% of concussion rate (Clarke, 1991).

Similar to other contact team sports, ice hockey experiences a high rate of head injuries. Hayes (1972) studied the number of the injuries in 560 team games during 1970-71 seasons. He found that 45% of the total incidences were to the head and face. In the head injuries ten concussion were reported. Dick (1993) reviewed the prevalence of concussions from 1986 to 1990 in ice hockey. He reported that the average rate of concussion was 4.2% of total number of incidence. The occurrence of brain injury in ice hockey from 1966 to 1997 has been reviewed by Honey (1998). Honey came to the conclusion that concussions were the most common brain injury in ice hockey. It increases with the level of the play; from 0.0 to 2.8 (per 1000 player hours) for players aged 5 to 14 years and for players aged on elite team the rate was 0.0-6.6. Honey reported that the concussion rate in elite levels is higher in game than practice. Hockey has the lowest rate (34%) of the injuries during practice and the highest rate (66%) during game (Dick, 1993). Using helmets is reduced the concussion rate in ice hockey; i.e. from 8.3% in 1968 to 3.8% in 1969 during game reported by Kraus in 1970 (Honey 1998). However, in contact sports, it is virtually impossible to protect the brain from impulse inertial loading, by means of a protective headgear (Gennarelli, 1991a).

2.3. Helmets

Helmets cushion the blow to the head and spread the force over a large area (Hurt and Thom, 1985). A helmet usually has two basic elements: an outer shell and an inner liner. Shells tend to be rigid, tough, and hard. On the other hand, the liner must have relatively constant low crushing strength, insensitive to the strain rate, and plastic in its crushing behavior. According to Newman (1980), maximizing energy absorption and minimizing the force development during head impact can be obtained by increased padding thickness and padding area, decreasing the crushing strength of the padding, and uniform crushing strength (Melvin et al., 1993). The engineering challenge is to design helmets with optimal shape and material to achieve the following goals: compress all the available liner material up to 80% of their thickness, eliminate rebound, maximize the onset and finishing rate, and maintain constant acceleration during impact.

Several studies have focused on understanding the relation between helmet design and head protection in sport. For instance, Bishop and Briard (1984) used a drop system to evaluate impact performance of bicycle helmets. Seven brands of bicycle helmets were impacted on four locations: front, rear, and left side from 1.0m and right front boss from 1.0m to 1.75m heights. A triaxial accelerometer mounted at the center of the gravity of a Hodgson-WSU headform recorded the shock (deceleration) that the helmets received. Acceleration signals were sampled at 6060.6 Hz and the yield peak acceleration (g peak) and GSI were calculated. Significant differences within and between the helmets were found. Front and rear locations provided better protection than side. Helmets with polystyrene liners, on first impact provided better protection than those with soft foam liners. Average GSI values were in the range of 800-1200 N.s., though two models were above 1500.

Another study by Bishop and Arnold (1993) determined the effectiveness of hockey helmets in limiting local loading on the side of the head. Their findings indicate that a risk of localized trauma at the temporoparietal region could exist from impacts with smaller magnitude than SI threshold. Six brands of hockey helmets were tested. Each helmet was mounted on a Hodgson-WSU headform, which could rotate freely about a vertical flexible post. Medium-density Fuji-film was located between headform and helmet at the left temporal area to measure peak headform pressure. Peak head acceleration was recorded by means of a triaxial accelerometer with 6600 Hz sampling frequency. Data from the Fuji film scanned by using light sensitive wand. To have more accurate result, the wand was connected to a Watscope A/D converter (sampling at 500 Hz) via a light densitometer. Each helmet was subjected to a single puck impact on the left side at 27.8, 33.4, and 38.9 m/s (100, 120, and 140 km/h). The peak pressures measured ranged from 12-33.4 MPa. This range was noted to be higher than the fracture tolerance of the temporoparietal area of the head (3.1 MPa). However, peak headform accelerations were from 250 to 275 g and Severity Indexes were all less than 800 that showed. Hence, SI and g values were not useful to conclude the severity of local injuries caused by a puck. The researchers suggested that helmet padding in this region should be improved, but future research was needed because their method did not provide any information regarding the time history and spatial distribution pattern of the applied pressure.

To quantify the magnitude of pressure and head acceleration due to side impact, Gilchrist and Mills (1996) conducted a unique study. Helmeted headforms mounted on a flexible neck, and torso were used to evaluate the importance of the neck and torso in direct impacts to the head. The flexible neck was found to protect the head from higher kinetic energy than fixed headform. The normal force was applied to the side of the ATD head by means of two flat and hemispherical strikers. A linear triaxial accelerometer with a cutoff frequency of 2000 Hz and angular velocity transducer with bandwidth of 0.3-1000 Hz were mounted at the center of the gravity of the headform to measure peak headform accelerations. Super low type Fuji film was placed under the helmet at the impact side. Other accelerometers where mounted at the torso-cervical junction. For the flat striker, they found that initial impact lasted about 5 ms while the headform rotated relative to the torso (50ms), which caused relative rotation of helmet to the headform. A second smaller impact, due to striker's rebound, occurred 50 ms after the first impact. The linear acceleration measured was about 100 g. F_{head} was found to equal to 90% of $F_{stricker}$. For the 35 mm diameter hemispherical striker, the first impact was perpendicular to the side of the helmet making a small tangential motion of the soft shell helmet. The neck controlled the helmet's rebound from the striker. The second impact between the striker and the rotated headform was oblique. Linear acceleration for the flat striker was more pronounced, but the peak was longer for hemispherical impactor. The shell deformation for the flat impactor was less than for hemispherical. They concluded that: the rotation of the helmet during impact caused more volume of the foam to crush; the real crushed area was underestimated by using a fixed headform; non-shell bike helmets can protect the head from impact energies between 30-60 J; foam thickness must be at least 20 mm, and the load spreading of soft shells were better than polystyrene foams. The Fuji-film was used to define the impacts site.

Different studies have performed to find the head acceleration limit, the thickness of padding liner, hardness of shell for different type of helmets, geometry and shape of the outer shell; however, the researchers focus was on determining peak linear accelerations and peak local pressures only and did not attempt to determine the life expectancy of the helmet. Players use a helmet for years in their professional life without knowing about its limited efficiency on protecting their head, because there is no warning sign for the duration of use on the helmets. Standards only test the helmets for the pass-fail criteria using the head linear acceleration, not the helmet's tolerance for repeated impacts.

2.3.1. Standards for Helmet Testing

Standards for sports helmets have evolved over several years. The fundamental functional properties of head protection were first defined by Lombard in the early 1950's (Hurt and Thom, 1985) when the modern helmet and face mask were introduced in football (Clarke, 1991). In the early 1960's, Swedish Ice Hockey Association established the standard testing method for helmets, which was dropping a weight into a fixed dummy head (Johnson, 1989). In the late 1960s the NOCSAE started to develop standards for all football helmets by using drop assembly. The aim was to test all helmets before being marketed (Clarke, 1991). In 1973, the Canadian Standards Association (CSA) published its first preliminary standard for hockey helmet (Dixon, 1989). The American Society for Testing and Material (ASTM) established its first standards for ice hockey helmet in 1975 when CSA and Swedish standard helmets were widely used by many players (Morehouse, 1989). In 1979, the need for international standard for ice hockey equipment is identified and the first standard of the International Organization for Standards (ISO) for hockey helmet and face mask was published in 1987 in Ottawa (Dixon and Brodie, 1993). Since there have been some doubt about reliability and validity of these standards, numerous research have been conducted to evaluate the effectiveness of these standards.The main goal of all standards is to minimize the risk of injuries in specific activity. For practical and ethical reasons human models and cadaver could not be used routinely in head injury studies; therefore, test protocols use surrogate anthropomorphic test forms.

For the ice hockey helmet, test parameters for certification are based on linear acceleration measures of a headform plus helmet during controlled impacts. Fail or pass criteria are established by these measures with respect to predefined performance thresholds. However, between standards the impact tests vary due to differences in drop apparatus, headforms, impact energy, pass/fail criteria, impact velocity, impact surface, and locations of impacts on the helmet (Dixon and Brodie, 1993; Pearsall et al., 2000; Halsted et al., 2000; Bishop, 1993; Pearsall et al., 2000; Sabelli and Morehouse, 2000). Some of these various aspects will be discussed in further detail in the following sections.

2.3.1.1. Head Form

Helmet standards use different headforms in their performance tests; some of them specified the headform, while others not. The magnesium-aluminum alloy headform (ASTM) is 5.11 kg. It has good tolerance to drop test on non-padded, rigid surfaces. It can be replaced and reproduced easily. On the other hand, the epoxy headform is only 0.12 g more than Magnesium headform, but it is susceptible to crack and hard to reproduce. The headforms accelerated differently for the same impact energy (Bishop 1993, Pearsall et al. 2000). The average peak accelerations were 335 and 396 g for epoxy and magnesium head forms respectively (Bishop 1993). Both headforms also showed different acceleration values on six test locations. The epoxy headform showed less rigid performance. Bishop (1993) concluded that there is no uniformity in performance of these headforms even within the headform. In the recent study, Bishop (2000) again reported that the magnesium head form shows higher peak acceleration than epoxy except on frontal impacts. However, based on linear regression model, Bishop concluded that a peak acceleration of 275 g for epoxy headform in CSA standards could be accepted for the peak value of 272 g of magnesium headform. And two headforms have almost same performances, if the other parameters stay constant.

Pearsall et al. (2000) noted the values of 175.7 g and 164.6 g for epoxy and magnesium headforms in helmet drop tests, respectively. They explained that the differences could be due in part to weight differences between headforms and their assemblies. They found some difficulty in locating rear boss on epoxy headform for CSA standard. Pearsall et al. (2000) also reported that using full-face headform for ISO standards alters the helmet fitting due to 27.5 mm difference in circumference of J and M size headform and consequently can alter the peak acceleration.

In summary, the material properties of headforms can influence test results substantially. Though metal alloy headforms do not posses bifidelity with respect to real head and brain tissue properties, by default they are the preferred material to perform testing repeatability.

2.3.1.2. Guide Assembly

Each standard uses different rigs for their performances, monorail, guide wire, or free fall. Bishop (1993) reported that monorail shows the peak acceleration of 31% higher than guide wire drop assembly in same impact energies. A part of difference could be due to friction and a great part could be due to eccentric loading of the headform. The eccentric loading helps to dissipate energy in both systems, but bending of the guide wire decreases energy more and the system produces less acceleration. The monorail system reaches its highest peak acceleration earlier. It means that this system can induce higher impact frequency and consequently high stiffness. As a result, if a helmet is passed in guide wire system, it could fail in monorail system. It could create a problem for helmet pass/fail criteria. By making stiffer helmets to pass the criteria of monorail system, it is possible that to create some problem for users (Bishop 1993). The velocity sensor could fall off during a test in a monorail system which can alter the results between the three trials (Pearsall et al. 2000).

2.3.1.3. Impact surface

Other difference between standards is the impact surface. ASTM standards uses modular elastic programmer (MEP), mounted on an aluminum plate, as impact surface, while others use steel surface on cement base (Pearsall et al. 2000). This difference may explain the overall impact peck acceleration for the ASTM standard that is lower than the other standards (Pearsall et al. 2000), while the impact energy is higher for ASTM standards (51 J versus 40 J for others).

On the other hand, ASTM standards use different velocity than others, 4.50 m/s versus 3.96 m/s for other standards. This stems from the fact that because MEP is less stiff than steel; therefore, more velocity is required. There are some problems in the long term use of an elastomeric surface due to some changes in its mechanical characteristics over time. Long time expose to ultra-violet light can change its hardness. Repeated mechanical use can damage elastomeric surface and consequently alter the result. Its reproducibility between and within manufacturers is difficult. Moreover, elastomeric surface has low impedance almost equal with test specimen, so can cause specimen-geometry dependency in the system. Because of all those problems, some degree of error will occur over time and between the test apparatus (Sabelli & Morehouse 2000).

In ambient condition, peck acceleration for impact at side and rear boss locations of helmet are less for elastomeric surface than steel surface. For high temperature conditions ($50\pm2^{\circ c}$), the peak accelerations for all locations were less for elastomeric surface (Sabelli and Morehouse 2000). Finally, there is temperature dependency of some helmet liners, especially in temperature between ambient and high temperature, during testing on elastomeric surface (Sabelli and Morehouse 2000).

2.3.1.4. Impact Velocity

ASTM standards use different velocity than others, 4.50 m/s versus 3.96 m/s for other standards. Also, the headform plus assembly mass is higher for ASTM standards, 6.1 kg versus 5.2 (Pearsall et al. 2000). High velocity and more mass will induce more impact kinetic energy ($E=\frac{1}{2}mv^2$) and consequently, high acceleration. Pearsall et al (2000) calculated this extra energy (0.74 J) and found that this energy can induce extra acceleration of 58.07 g. The pass/fail criteria is 300 g, so they concluded that the extra energy could be absorbed with a less rigid elastomeric surface.

The main goal of all standards is to test efficiency of helmets in protecting head from injuries. Similar activities should have similar test standards. Differences between standards can create variation in pass/fail criteria of the similar products. The standards have varieties in impact energy, impact velocity, impact surface, drop mass, guide assembly, headforms, and definition of impact locations and reference plane on the helmet. All these varieties can cause misleading results and the results from one system cannot be compared with the results from other systems (Bishop 1993).

There are some additional parameters that are not included in some standards. For instance, CSA standards do not include high temperature conditions, however during the game, the temperature of the inside of the helmet can be up to 41-43°c (Halstead et al. 2000). All standards test blunt impacts on flat surfaces and there is no test for high-velocity/low-mass focal impacts or to simulate other conditions, i.e., head impact with skate blade. Facemask and strap are connected to the helmet and they could alter the helmet performance, therefore, standards have to measure the exerting force (Halstead et al. 2000).

Pearsall et al. (2000) compared the four different hockey helmets with four different standards for testing hockey helmet including: ASRM, CEN, CSA, and ISO standards. They did not find statistical differences between standards when comparing peak helmet plus headform acceleration. By considering the impact site and helmet model, they found significant differences ($p<0.05$) between standards. They concluded, "there was not one standard that stood out as being more or less severe, given that each site showed different trends; that is, there were no main effects evident."

3. Methods

The following sections outline the testing apparatus, protocol and research design implemented in this study.

3.1. Apparatus

A monorail drop testing apparatus in the Product Testing Laboratory of Bauer-Nike Hockey Inc. (Saint-Jerome, Quebec, Canada) was used to conduct controlled impact tests. The drop testing apparatus permits helmet testing according to the CSA standard CAN/CSA-Z262.1-M90 (re-adopted in 1995). In conjunction with this apparatus, a uniaxial linear piezoelectric accelerometer (Model 353B04, ± 500 g, Dalimar Instruments Inc.) at the center of the gravity of the headform (ISO, large size, M) measured the peak impact linear deceleration. The drop height was adjusted to obtain desired impact kinetic energy levels. A photoelectric cell located on the fixed post was used to confirm velocity of free fall 40 mm prior to the impact and thus confirm kinetic energy. The CADEX^{TM} software (St. Jean Sur Richelieu, QC) was used to record and condition the output signals (frequency range 0 to 1000 Hz With a ±1.5 % variance; sampling rate 10 KHz; filtering acquisition use 1000 Hz). All tests were conducted at ambient room temperature (20 ± 2°C) and relative humidity of 55%. The impact surface was a flat steel anvil that was supported on a rigid, concrete foundation (Figure 3).

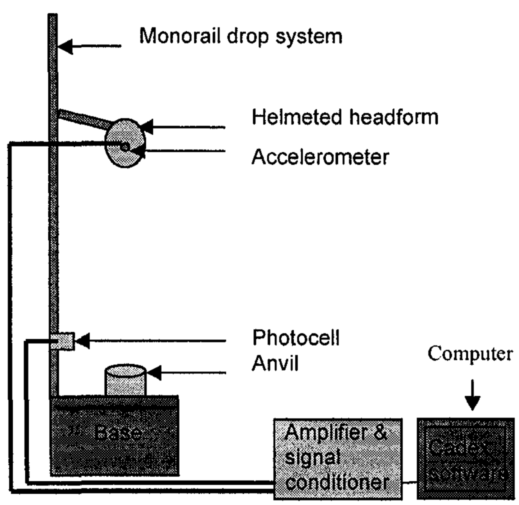


Figure 3: Testing apparatus. Helmet, headform and accelerometer are linked to vertically sliding armature to guide the drop and impact onto the anvil. Analog signals from the accelerometer and photocell are collected and record to computer.

3.2. Testing Protocol

A series of pilot studies were conducted using three different energy levels (30, 40, and 50 J) to determine the range of impact energies to use for actual testing. This data showed that all helmets under consideration could successfully sustain 50 repeated impacts per site at the 30 J without changes to attenuation properties; therefore, for the subsequent complete study, tests were conducted at only the higher impact energies of 40, and 50 J, plus 60 J. With the combined headform and carriage mass of 5.2 kg, the heights for impacts were 84, 104, and 124 centimeters for 40, 50, and 60 J respectively. The measurements were from the top of the anvil to the bottom of the helmet.

Five models of helmet were used: Bauer-Nike HH3000, HH4000, HH5000, Jofa XHT690, and CCM HT500 (figure 4). The first three models were all white and the last two models were black. All helmets were large size (i.e. head transverse circumference 550 mm). A total of 45 samples were tested (3 samples per each energy level from each helmet model).

HH3000 JOFA-XHT690 HH5000 CCM-HT500 HH4000

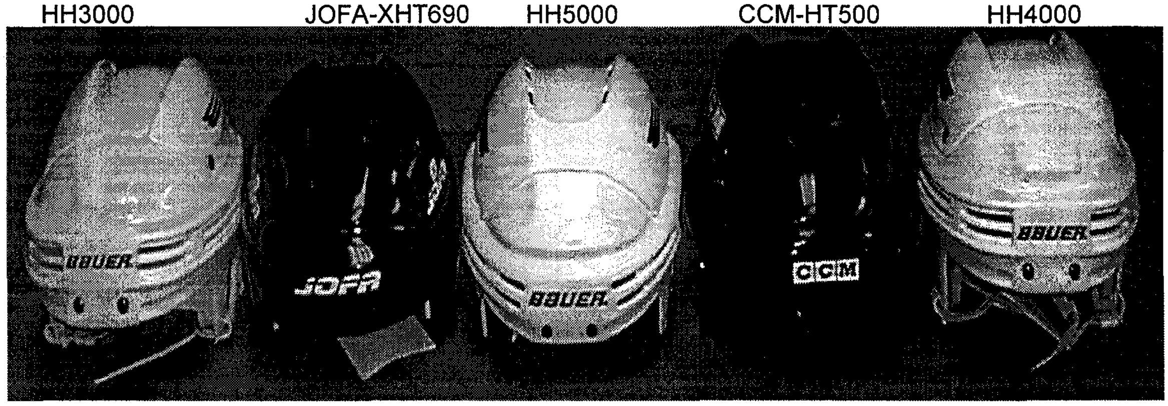


Figure 4: The ice hockey helmet models used in the study.

The foams used for the liners of the helmets were Expanded Polypropylene (EPP), Vinyl Nitrile (VN, brandnames Cellflex®, and Rubatex®) padding. VN foam with the average thickness of 16.5 mm was used in HH3000. In the HH4000 helmets dual combined black high modulus VN foam (4 mm) positioned against the shell and optimal modulus VN foam (12.5 mm at the front, rear, and crown; 9 mm at the sides) positioned against the head. EPP foam was used in HH5000, CCM, and JOFA helmets. EPP was covered with different comfort padding to eliminate pressure points and grip the head. VN pads were used in CCM helmets and Rubatex® pads were used in JOFA helmets (table 4).

Table 4: Different Liners and padding used for the helmets

Foam/Model	CCM	HH3000	HH4000	HH5000	JOFA
Liner	EPP	VN	Dual VN	EPP	EPP
Comfort pad	VN	--	--	PVC	Rubatex

All the tags were removed from the helmets. Helmets were labeled on the exterior surface of the shell with a grease pen indicating impact energy level (40, 50, or 60J) and sample number (1, 2, or 3). The helmets were randomly mounted on an ISO large size adult headform that in turn was fixed on drop assembly (i.e combined mass = 5.2 kg). To improve mounting and prevent helmet slippage during impact, the chin straps were secured taunt around the headform. All helmets were impact tested in 4 different locations (i.e. front, right side, back, and crown) according to CAN/CSA-Z262.1-M90 guidelines (Figure 5).

Each location on the helmets was impacted 50 times at one of the levels of kinetic energy. Secondary impacts after rebounding from the principal impact were not allowed. After each impact, the helmet was relocated on the headform to ensure the same impact site occurred for all 50 impacts. To visually confirm the helmet impact point, the surface of the anvil was coated with color transferring paint. Each helmet received a total of 200 impacts. The time interval between multiple impacts was 30 to 60 seconds as according to CAN/CSA-Z262.1-M90.

The peak headform accelerations, impact velocity and SI-index were recorded for each impact; moreover, helmets were evaluated visually to document any shell crack/deformation or liner crushing/tearing. The result from each trial was compared with the other trials and performance of all types of helmets.

The impact tests were stopped when 5 or more consecutive "fails" i.e. greater than 275 g) occurred and/or the recorded peak acceleration reached 400 g (i.e. the acceleration limit to protect the drop system from damage).

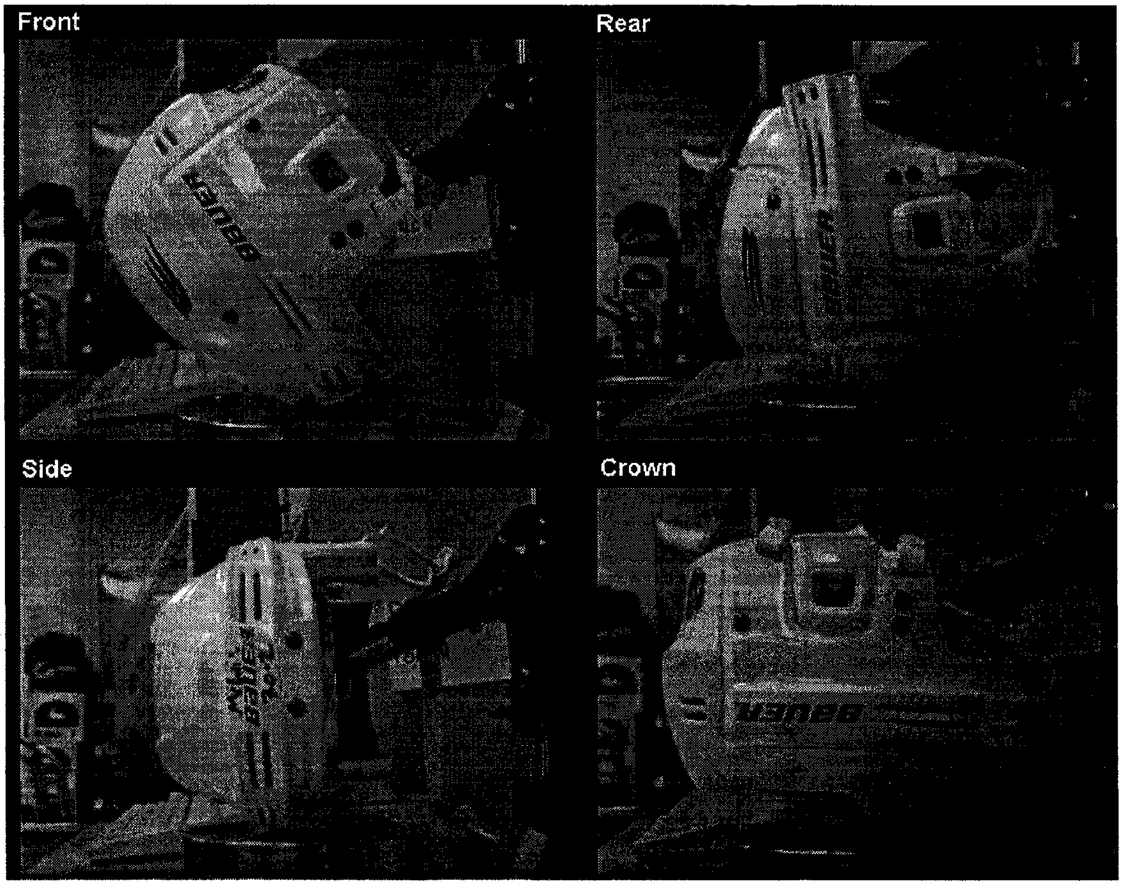


Figure 5: Helmet impact locations: front, rear, side, and crown

3.3. Research Design

This study has identified three factors for examination: helmet model, impact site and impact number (the latter being a repeated measure). Two independent variables were recorded from each test: peak G and severity index (SI) score. Hence the research design can be summarized as follows:

$$S_n \text{ x } (HM_n \text{ x } E_n \text{ x } Site_n \text{ x } I_n)$$

where

S_n = sample number (1 of 3)

HM_n = helmet model

(1 of 5: BNH HH3000, HH4000, HH5000, CCM HT500, JOFA XTH 690)

E_n = energy of impact (40, 50, or 60 J)

$Site_n$ = impact site (1 of 4: front, rear, side, and crown)

I_n = impact number (1 of 50)

In total, 9000 unique impact trials were conducted. Excel® (Microsoft® Excel 2002) spreadsheets were used to combine and organize data files, while Statistica® (StatSoft, Inc., 5.0) was used for statistical evaluation of the results.

Assessment of the results involved: (1) description of performance of helmets by sites with respect to "pass-fail" criteria (i.e. peak G threshold of 275 g and SI threshold of 1500) and the corresponding multiple impact limit; (2) creation of impact performance profile for each helmet by site i.e. generation of regression functions to describe changes in attenuation properties with respect to multiple impacts; (3) comparison of impact measures of the 1st, 2nd, 3rd, 5th, 10th, 20th, 30th, 40th, and 50th impacts (where permissible) within each helmet model by site (1 way RM MANOVA); and, (4) comparison between helmet models by site for corresponding impact numbers (1 way ANOVA where helmet model is the sole factor e.g. 3rd impact of helmet A versus 3rd impact of helmet B for side impacts .

4. RESULTS

The results are subdivided into sections describing the general effect of the main factors on impact attenuation performance followed by a focus on the specific factor interactions.

4.1. Multiple impact performance thresholds

In general, all helmets were found to pass the CSA (CAN/CSA-Z262.1-M90) standards; however, beyond three repeated impacts and above 40 J, attenuation properties were substantially reduced. All the helmets demonstrated excellent attenuation properties at the crown, rear, and front for the energy level of 40 J; however, particularly poor attenuation properties beyond 40 J and three impacts were shown for side impacts. More specifically, side impacts at 40 J were well sustained for the models A, B, and C. At 50 J of impact energy, all helmets passed at the front and crown impacts; however, it was not possible to finish 50 repeated impacts for the models A and B at rear sites nor for all the models at the side impacts. At 60 J of impact energy, all the helmet models passed at the crown except the model C. The 50 repeated impacts were not completed for the model E (at the side site) and for the model D (at side and rear sites). Table 5 shows the average number of impacts that each model tolerated before failing. Some unexpected outcomes are worth noting: 1) for the helmet model C, the SI values (Severity Index) reached the threshold point earlier than the peak head acceleration (peak G) values, and, 2)the average number of impacts that the model A tolerated at the rear site with 50 J of energy was smaller than the 60 J.

Table 6 shows the average peak acceleration range for all the impacts. The minimum and the maximum values for all the sites and all the helmet models increased with increasing the energy level from 40 to 60 J.

Table 5: The average number of impacts until peak G > 275or SI> 1500

Energy	40 J				50 J				60 J			
Model / Site	C	F	R	S	C	F	R	S	C	F	R	S
A	✓	✓	✓	✓	✓	✓	**<u>18</u>**	12	✓	13	**<u>21</u>**	8
B	✓	✓	✓	48	✓	✓	30	6	✓	42	17	3
C	✓	✓	✓	✓	✓	✓	45	6 SI 10g	21SI 49g	26SI 43g	15SI 35g	2SI 3g
D	✓	✓	✓	5	✓	✓	✓	2	✓	✓	12	1-2
E	✓	✓	✓	4	✓	✓	✓	3	✓	✓	✓	2

* ✓ indicates that the 50 repeated impacts were completed successfully.
* g = average number of impacts with g > 275
* s = average number of impacts with SI >1500

Table 6: Average peak acceleration range (see the text for details)

Energy	Site/Model	A	B	C	D	E
40	Crown	102-146	104-144	137-190	105-130	116-138
	Front	89-184	100-141	125-200	105-130	108-128
	Rear	98-142	97-178	92-153	105-148	128-132
	Side	94-273	123-304	151-273	147-404	131-376
50	Crown	113-197	116-181	151-208	110-139	131-145
	Front	127-220	114-153	132-206	110-139	127-163
	Rear	107-289	104-291	93-216	123-203	134-148
	Side	118-323	154-337	186-302	181-424	153- 390
60	Crown	133-263	133-237	172-281	122-174	139-178
	Front	159-361	148-320	157-287	122-174	132- 214
	Rear	116-399	188-342	117-272	127-324	150-166
	Side	137-363	188-382	208-361	267 -559	240-442

From the above findings (table 5) it was evident that for some helmets the peak acceleration of headform was greater than 275 g even after a second impact; thus, it was impossible to complete 50 trials for all factor combinations. As a consequence the repeated measures yielded an uneven data set; therefore, subsequent results will be presented from general to more specific analyses including: 1) the first three repeated impacts, 2) fifty repeated impacts, 3) repeated impacts at 40J, 4) repeated impacts at 50J, 5) repeated impacts at 60J, and 6) peak G prediction functions.

4.2. First three repeated impacts

Preliminary analysis was conducted for the first three trials including all helmet models, sites, and energy levels (tables 7-11 and 12); there were significant main effects and interactions for all the factors at the first three impacts ($p<0.001$).

Table 12: ANOVA values for peak Gs including: helmets, sites, and the energy levels (up to the third impact)

1-MODEL, 2-SITE, 3-ENERGY, 4-TRIAL						
	Df Effect	MS Effect	df Error	MS Error	F	p-level
1	4	19648.04	118	519.25	37.84	0.00
2	3	253426.55	118	519.25	488.06	0.00
3	2	148696.64	118	519.25	286.37	0.00
4	2	103294.77	236	101.94	1013.29	0.00
12	12	23304.40	118	519.25	44.88	0.00
13	8	2413.33	118	519.25	4.65	0.00
23	6	14311.73	118	519.25	27.56	0.00
14	8	1776.01	236	101.94	17.42	0.00
24	6	22205.62	236	101.94	217.83	0.00
34	4	3593.67	236	101.94	35.25	0.00
123	24	3316.76	118	519.25	6.39	0.00
124	24	2574.36	236	101.94	25.25	0.00
134	16	233.99	236	101.94	2.30	0.00
234	12	846.95	236	101.94	8.31	0.00
1234	48	327.09	236	101.94	3.21	0.00

The post hoc test (Tukey) showed that the helmet models A and B were significantly different ($p<0.001$) than other models (figure 6); there was no significant difference between the other models. The comparison between the sites showed that the side impacts were significantly different ($p<0.001$) than other sites; there was no significant difference between the other sites (figure 7). There were significant differences ($p<0.001$) for all the energy levels and all the trials from 1st to the 3rd trial.

Table 7: Peak G Means and Standard Deviations for Helmet Model A.
The empty cells indicate that 50 impacts were not completed in that trial.

	Site	Energy										
	Crown			Front			Rear			Side		
Trial	40	50	60	40	50	60	40	50	60	40	50	60
1	102± 7	113± 6	133 ± 2	89 ± 5	127 ± 5	159 ± 1	98 ± 5	109±10	116 ± 8	94 ± 12	122 ± 5	137 ± 1
2	110± 4	131± 7	154 ± 2	128±16	154 ± 5	160±40	103 ± 6	122±22	140 ± 3	117±30	173±24	186±28
3	114± 5	139±10	161 ± 7	137±14	167 ± 4	198 ± 6	106 ± 3	140±29	156 ± 4	136±50	218±47	220±43
4	121± 6	138± 2	169±16	146 ± 9	171 ± 6	202±12	107 ± 3	152±36	180 ± 6	122±13	22±136	231±35
5	123±11	145±10	156 ± 9	153 ± 8	175 ± 9	215±12	109 ± 6	160±41	187±12	125±13	257±53	237±21
10	128±14	143± 4	173±10	155 ± 1	185±10	262±14	114 ± 9	219±51	221±28	150 ± 4	330±70	293±30
20	132±55	153± 8	201±26	167 ± 2	190 ± 8	360±30	124 ± 6	267±30	274±49	172±10		
30	134±22	182±28	197±34	168 ± 6	197±15	383±14	132 ±13	310±62	343±55	197±33		
40	127±16	178±20	232±47	180 ± 9	198±15	384±14	289±154	306±46	348±46	216±66		
50	135±14	171±14	231±12	170 ± 5	235±10	384±13	139±16	334±64	348±46	196±35		

Table 8: Peak G Means and Standard Deviations for Helmet Model B.
The empty cells indicate that 50 impacts were not completed in that trial.

	Site	Energy										
Trial	Crown			Front			Rear			Side		
1	104 ± 4	115 ± 2	133 ± 1	100±12	114 ± 7	148±1	97 ± 2	122±26	153±34	125 ± 5	154 ± 5	188±21
2	126 ± 4	133 ± 6	154 ± 4	117 ± 9	135 ± 5	160±3	105 ± 2	139±24	193±43	151 ± 7	193±15	234 ± 5
3	127±10	139 ± 5	158 ± 6	122 ± 4	139 ± 2	166±6	112 ± 2	137±35	242±86	168±10	224±28	290 ± 8
4	132 ± 8	140 ± 1	167 ± 1	127 ± 5	141 ± 3	161±3	117 ± 6	168±33	238±58	181±11	262±52	352±21
5	130±11	144 ± 6	169 ± 1	128 ± 1	144 ± 2	164±5	123 ± 8	208±64	230±55	181 ± 8	248±42	388±13
10	126 ± 3	145 ± 6	177 ± 7	150±25	145 ± 2	174±16	136±15	246±60	241±66	206±13	285±49	
20	129 ± 3	153±13	193±14	148±25	148 ± 1	233±25	155±21	261±67	322±42	254±32		
30	136±24	181±37	201±29	136 ± 3	149 ± 1	249±5	159±23	333±48	385±89	298±78		
40	140±20	161±22	208±30	133 ± 7	152 ± 7	295±28	176±31	370±54	385±89	281±82		
50	136±14	181±30	226±15	135 ± 4	150 ± 4	326±39	165±25	401±78	385±89	307±59		

Table 9: Peak G Means and Standard Deviations for Helmet Model C.
The empty cells indicate that 50 impacts were not completed in that trial.

	Site	Energy										
Trial	Crown			Front			Rear			Side		
1	139 ± 4	151 ± 6	173 ± 5	125 ± 9	132 ± 3	153 ± 7	99 ±12	91 ± 4	115 ± 5	157 ± 8	185 ± 7	208±11
2	158 ± 2	169±14	195 ± 4	130 ± 7	149 ± 2	189 ± 7	110 ± 8	116 ± 4	139 ± 6	216±29	227 ± 5	257±18
3	166 ± 5	170±16	203 ± 7	132 ± 9	161 ± 5	200±24	110±14	128 ± 7	159±10	228±33	234±19	287±14
4	167±14	164±16	209 ± 7	138 ± 6	170 ± 8	206±23	114±18	137 ± 9	173 ± 6	241±30	241±18	301±13
5	177±10	175±11	207±10	145 ± 2	170±15	213±17	115±19	145±12	190±16	248±25	263 ± 9	311±10
10	171±11	181±10	220±10	157±21	188 ± 8	228±15	125±21	191±44	251±31	267±28	284±13	
20	169±33	193 ± 8	231 ± 3	183±28	196±13	238±15	139±15	217±48	271 ± 8	275±25		
30	170±27	173±52	245±13	186±21	200 ± 8	253±11	141±21	214±52	269±11	260±10		
40	145±27	204 ± 1	269±23	191±23	203 ± 6	267±11	150±20	233±49	269±12	251±27		
50	156±25	202 ± 9	283±40	195±22	203 ± 5	296±19	152±21	249±60	265±17	253±30		

Table 10: Peak G Means and Standard Deviations for Helmet Model D.
The empty cells indicate that 50 impacts were not completed in that trial.

	Site	Energy										
Trial	40	50	60	40	50	60	40	50	60	40	50	60
1	94 ±4	105 ±3	118±12	94 ±10	110±3	122±3	105±14	123±10	126 ±2	121 ±4	181±21	267±23
2	110±5	129 ±3	144 ±5	107 ±2	123±3	138±4	114 ± 9	137 ±8	145 ±6	189 ±8	306±28	443±28
3	118±6	142 ±3	148±11	111 ±2	125±4	143±8	118 ± 7	146 ±9	169 ±7	238±15	388±10	520±34
4	123±4	151 ±4	165±14	113 ±4	128±3	145±5	122 ± 8	154 ±6	212±53	260±13	421±10	
5	126±4	157 ±4	154±16	114 ±3	132±5	147±4	123 ± 7	158 ±7	213±34	284±17	450 ±1	
10	127±3	171 ±3	214 ±9	118 ±3	135±5	152±2	126±10	173±10	311±96	317±79		
20	137±3	179±20	261±30	116 ±3	135±2	161±6	129 ± 6	186±13	342±69	382±28		
30	141±6	199±16	358±33	119 ±7	137±6	167±5	135 ± 3	195±17	342±69			
40	144±2	237±21	369±46	119 ±6	137±1	171±4	148±15	196±13	342±69			
50	144±4	244 ±3	369±46	118 ±8	137±6	174±2	144±17	200±11	342±69			

Table 11: Peak G Means and Standard Deviations for Helmet Model E.
The empty cells indicate that 50 impacts were not completed in that trial.

	Site	Energy										
Trial	40	50	60	40	50	60	40	50	60	40	50	60
1	116±13	131 ±3	139±6	108±8	124±10	131 ±9	128±7	134 ±5	149 ±4	131±14	153±14	240 ± 6
2	127 ± 1	145±10	160±3	116±7	135 ±9	156 ±8	134±1	141±10	161 ±2	219±19	267±29	375±12
3	124 ± 3	142±15	156±4	120±4	141 ±7	166 ±9	135±3	137±10	160 ±6	259±36	320±35	441±17
4	125 ± 2	145±12	156±1	121±5	143±10	172 ±9	135±5	148 ±5	158 ±3	297 ±6	336±43	
5	125 ± 5	146±16	156±1	121±6	144±10	175±11	129±6	136 ±7	160 ±2	311 ±6	350±31	
10	124 ± 3	141±13	161±4	123±4	149 ± 7	182±13	130±6	146 ±9	154 ±7	345±28	381 ±8	
20	128 ± 4	140±11	165±5	123±2	154 ± 9	199±12	129±5	143±11	160 ±3			
30	127 ±10	141 ±9	163±9	123±6	155 ± 4	203±10	126±6	145±10	166±12			
40	132 ± 2	144 ±6	168±6	121±5	161 ±3	212 ±5	127±1	144±11	160 ±8			
50	140 ± 8	139 ±6	174±2	126±1	159±10	213 ±9	128±5	146 ±9	164±11			

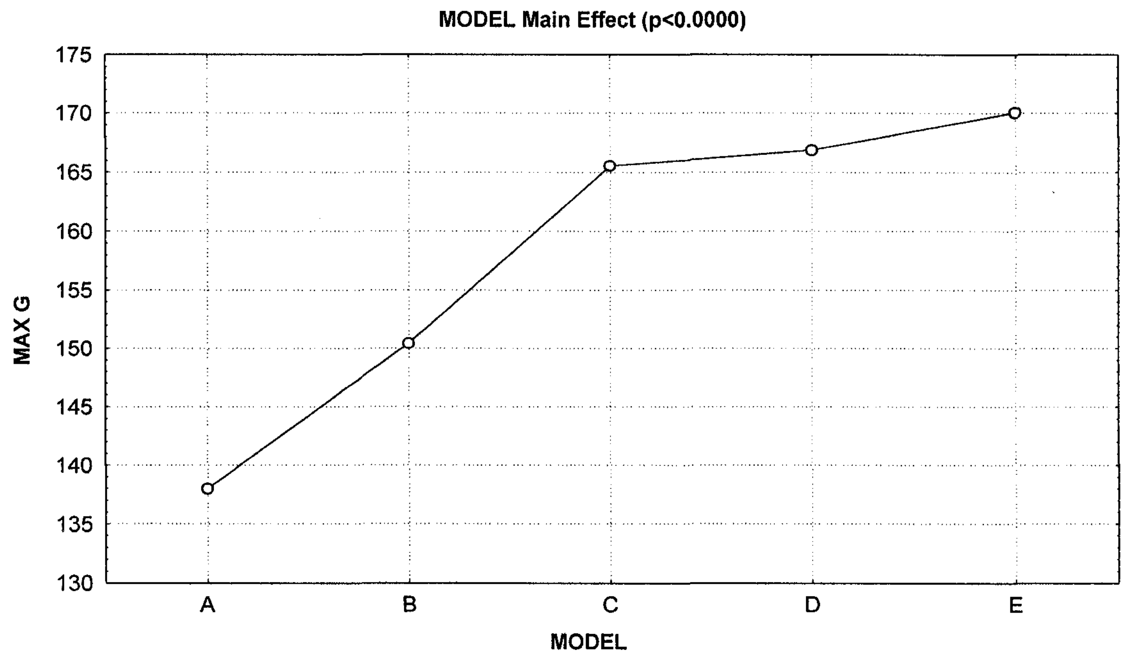


Figure 6: Average peak G for the helmet models at all the sites and all the energy levels. The models A and B were significantly different ($p<0.001$) from C, D, and E.

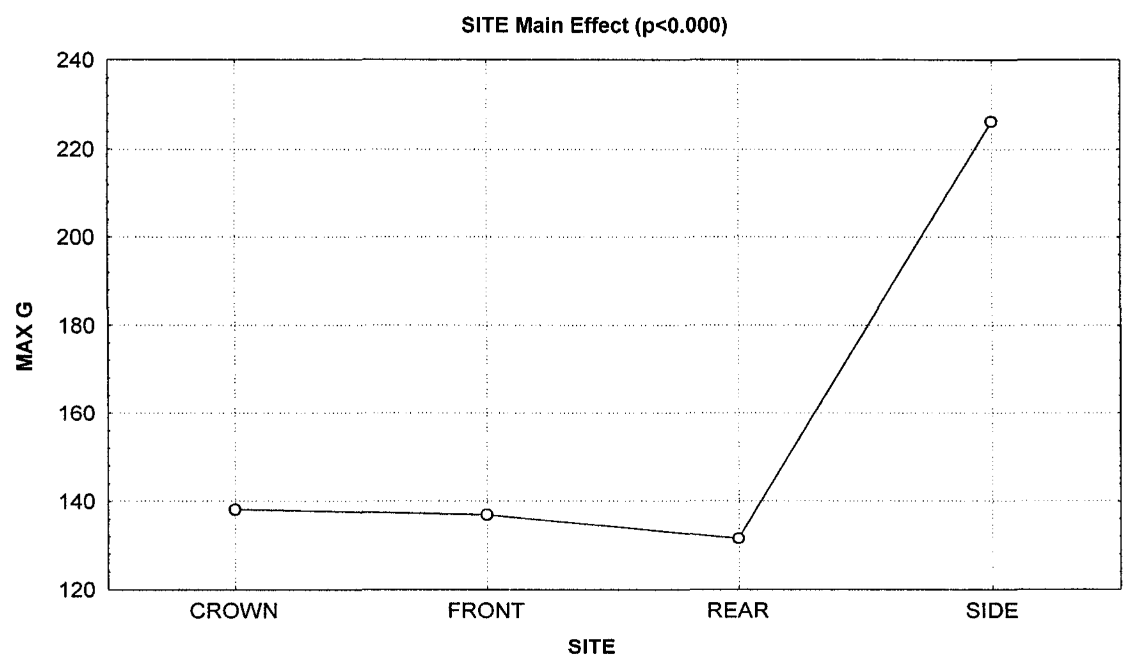


Figure 7: Average peak G for the different sites for all the helmets and the energies. The side impacts were significantly different ($p<0.001$) from the other three sites.

Figure 8 show there were significant differences (p<0.001) between the side impacts and the other sites. All sites showed significant difference for the different energy levels from 40 to 50 and 60 J, except for the crown impacts between the 40 and the 50 J of energy.

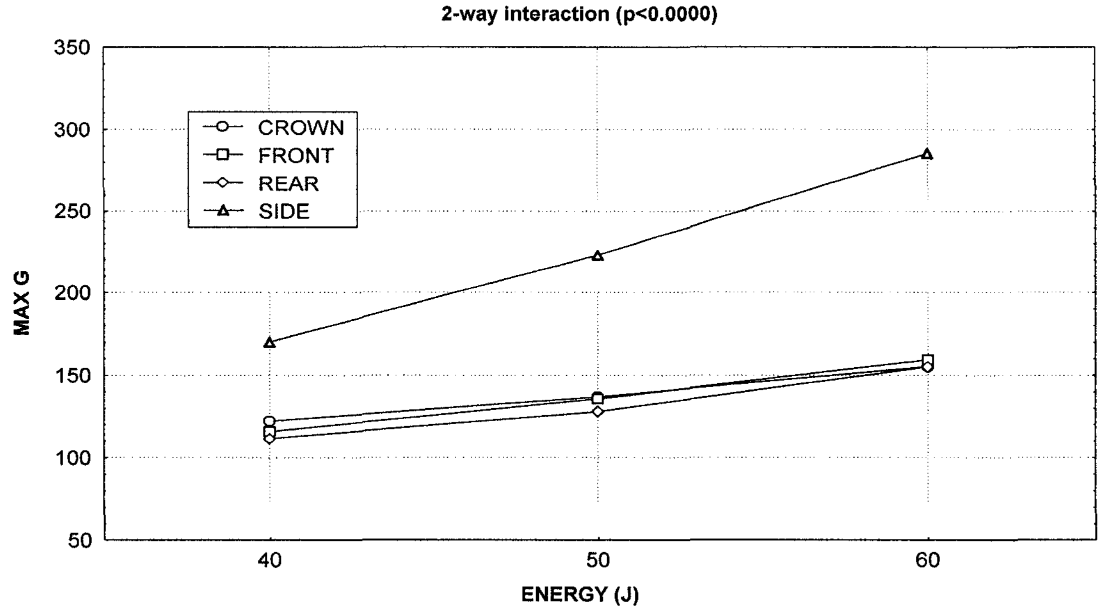


Figure 8: Average peak G for all the sites and energies for all the helmets. There were significant differences (p<0.001) for all the sites from one energy level to the next. The side impacts were significantly different (p<0.001) from the other three sites.

Figure 9 shows significant differences (p<0.001) between the side impacts and the other sites for all the helmet models; except for the model A, there was no significant deference between side and front impacts. The crown impacts for the C model showed significant difference (p<0.001) with the other models. There were significant differences (p<0.001) between front impacts for the models D with A and C. There were significant differences (p<0.001) between rear impacts for the models B with A and C, also C with E.

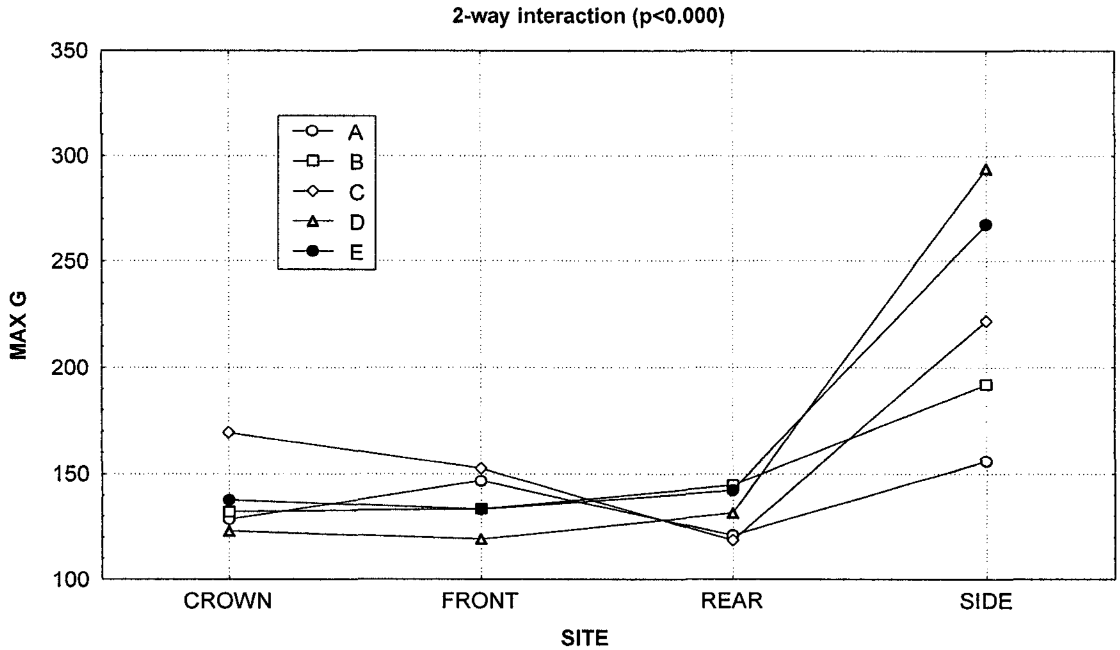


Figure 9: Comparison of peak G between 4 sites for all the helmet models
(see the text for details)

Figure 10 shows that there were significant differences ($p<0.001$) between the three energy levels for all the helmets, except for the model C between the energy of 40 and 50 J. There were some significant differences ($p<0.001$) between the models at the 40 J energy level: the model A with the models C and E; the model B with the models C and E; the model C with the models A, B, and D; the model D with the model C; the model E with the models A and B.

Comparison between 3 energy levels showed that there were some significant differences ($p<0.001$) between the models at the energy level of 50 J: the model A with the models D and E; the model B with the models C, D, and E; the model D with the model A and B; the model E with the models A and B. There were no significant differences between the models C with the other models (figure 10).

Moreover, there were some significant differences ($p<0.001$) between the models at the 60 J energy level: the model A with all the models; the model B with the models A and D; the model C with the model A; the model D with the models A and B; the model E with the model A.

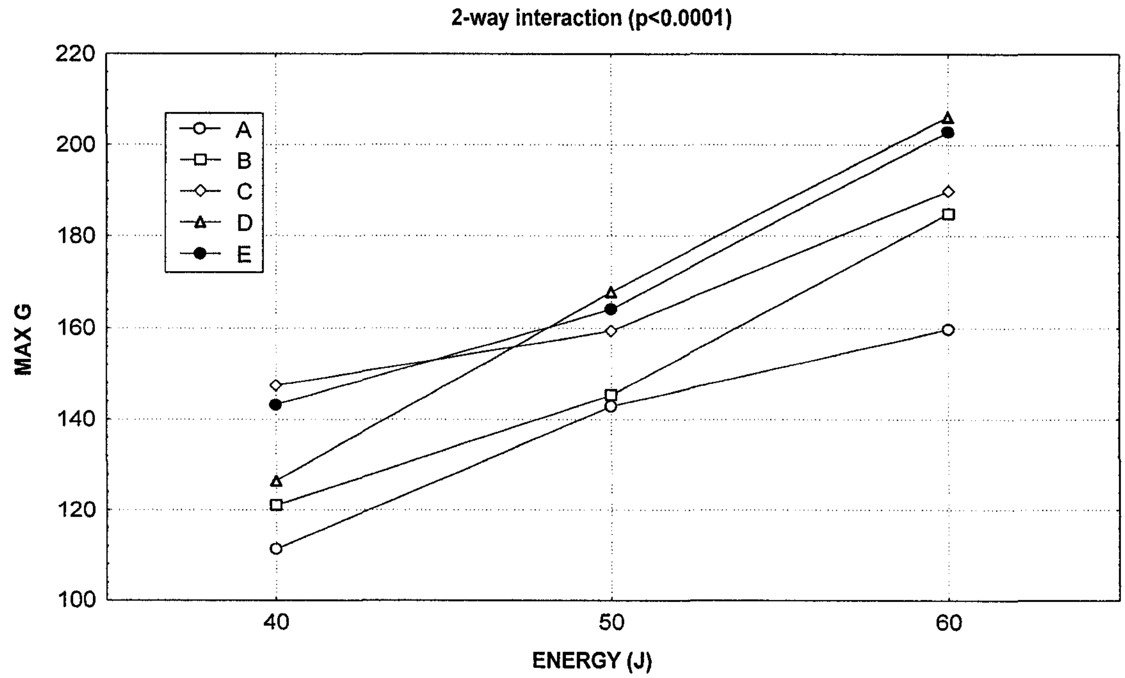


Figure 10: Comparison of peak G between 3 energy levels for all the helmet models (all sites combined)

Three way comparison between the 3 trials, 4 sites and 3 energy levels showed that there were significant interaction between trials and the energy levels; except, there was no interaction between all the 3 trials at the 40 J and the trial 1 and 2 at the 50 J for the rear impacts (figure 11). Side impacts were significantly different ($p<0.001$) from the other sites for all three trials at all the energy levels; except, there was not significant difference between side and crown impact at 40 J for the first trial. There were no significant differences between the crown, rear, and front impacts at all 3 trials and all the energy levels.

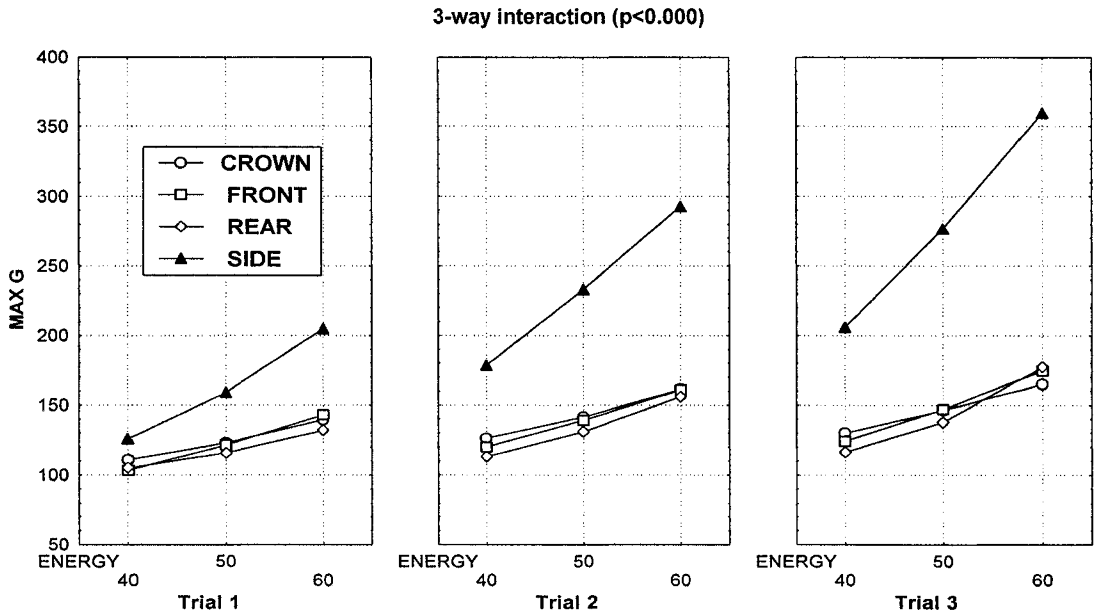


Figure 11: Comparison of peak G between the 3 trials, 4 sites and 3 energy levels (helmet models pooled). There were significant interactions (p<0.001) between all the trials and energy levels (see the text for more details).

4.3. Fifty repeated impacts

With the exception of the side locations, it was possible to evaluate the peak acceleration attenuation properties for the 50 repeated impacts for all the helmet models and energy levels (table 7-11 and 13). Table 13 shows significant main effects and interactions for all the factors (p< 0.001).

Different post hoc tests (Tukey test) conducted to find the significant differences; there was significant differences (p<0.001) between all the energy levels from 40 to 50 and 60 J. The test for the helmet model showed the models D and E were significantly different (p<0.001) from the other models; there was no significant difference between the other models.

Comparison test between the helmet models and the sites showed that there were some significant differences (p<0.001) between the helmets and the sites for 50 impacts. For the crown impacts, the means for the models A and B were significantly difference (p<0.001) from the mean of C; D with E, C with A, B, and E, and E with C and D. For the front impacts, there were no significant differences between the model A and C, D and E, and E and B. For the rear impacts, there were some significant differences ($p<0.001$) between the means: A with E; B with C, D, and E; C with B and E; D with B and E; E with all the other models.

Table 13: ANOVA values for peak Gs including: helmets, trials, energy levels, and sites (except side impacts).

1-MODEL, 2-SITE, 3-ENERGY, 4-RIALS						
	Df	MS	df	MS		
	Effect	Effect	Error	Error	F	p-level
1	4	63514.98	90	2372.50	26.77	0.00
2	2	45648.16	90	2372.50	19.24	0.00
3	2	671138.94	90	2372.50	282.88	0.00
4	9	132315.50	810	324.30	408.01	0.00
12	8	52261.38	90	2372.50	22.03	0.00
13	8	11621.45	90	2372.50	4.90	0.00
23	4	19392.98	90	2372.50	8.17	0.00
14	36	6565.42	810	324.30	20.25	0.00
24	18	7418.90	810	324.30	22.88	0.00
34	18	15844.76	810	324.30	48.86	0.00
123	16	11217.24	90	2372.50	4.73	0.00
124	72	3395.07	810	324.30	10.47	0.00
134	72	1322.11	810	324.30	4.08	0.00
234	36	1873.63	810	324.30	5.78	0.00
1234	144	1306.61	810	324.30	4.03	0.00

Figure 12 shows that the 2nd, 3rd, 4th, 5th, 10th, 20th, 30th, 40th, and 50th impacts were significantly different ($p<0.00$) from the 1st impact. There were significant differences ($p<0.001$) between all the trials, except between 4th with 3rd and 5th impact and trial 40th with the 50th.

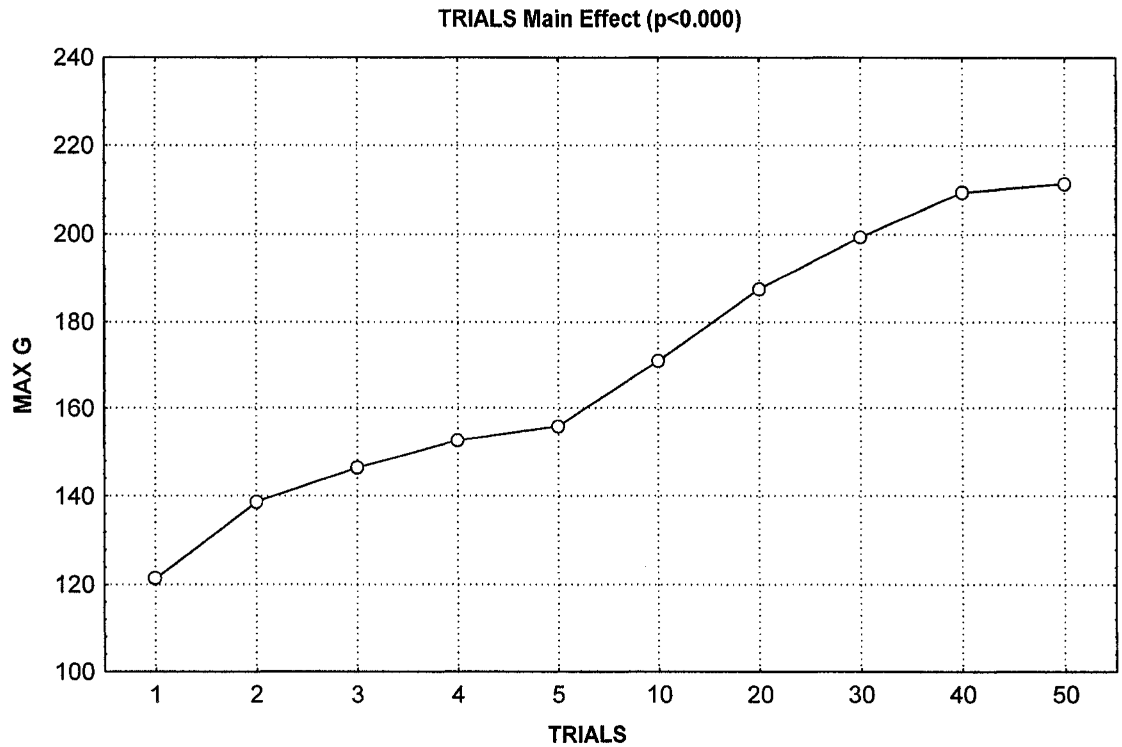


Figure 12: Average peak -G for all the trials.
All the trials were significantly different ($p<0.00$) from the 1st trial.

Figure 13 shows that for all the models, there were significant differences ($p<0.001$) between all trials with the first trial, except the model E; only trials 20, 30, 40 and 50 were significantly different from the first trial. There were significant differences ($p<0.001$) between the trials; however, there was no significant difference between each trial with its previous and the next trial. For instance, the 4th trial had no significant difference with the 3rd and 5th trial. There was no significant difference between the trials for the model E. this model has significant difference ($p<0.001$) with the other models after the 5th impact.

Figure 14 shows that there were significant differences ($p<0.001$) between all trials with the first trial and other trials; however, there was no significant difference between each trial with its previous and the next trial. The rear impacts were significantly different ($p<0.001$) from the crown and front impacts after the 5th impact; there was no statistical difference between crown and front impacts.

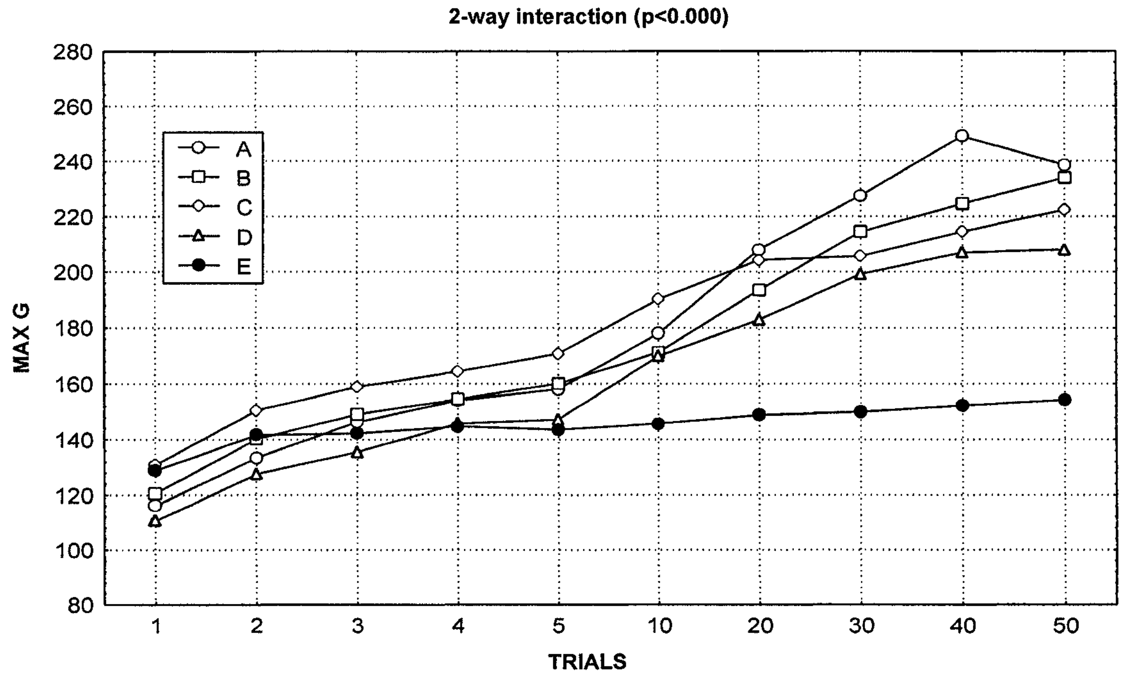


Figure 13: Average peak Gs for the 50 trials for different helmet models. The model E was significantly different (p<0.001) from the other model (see the text for details).

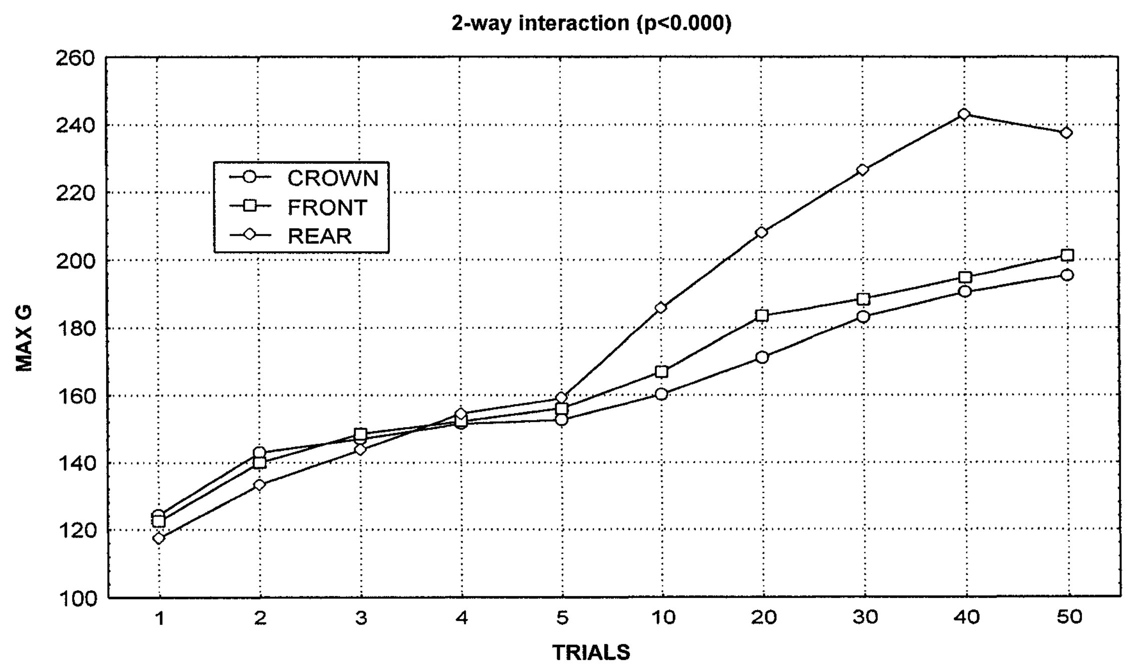


Figure 14: Average peak Gs for the 50 trials, comparison between the sites and the trials (see the text for the details).

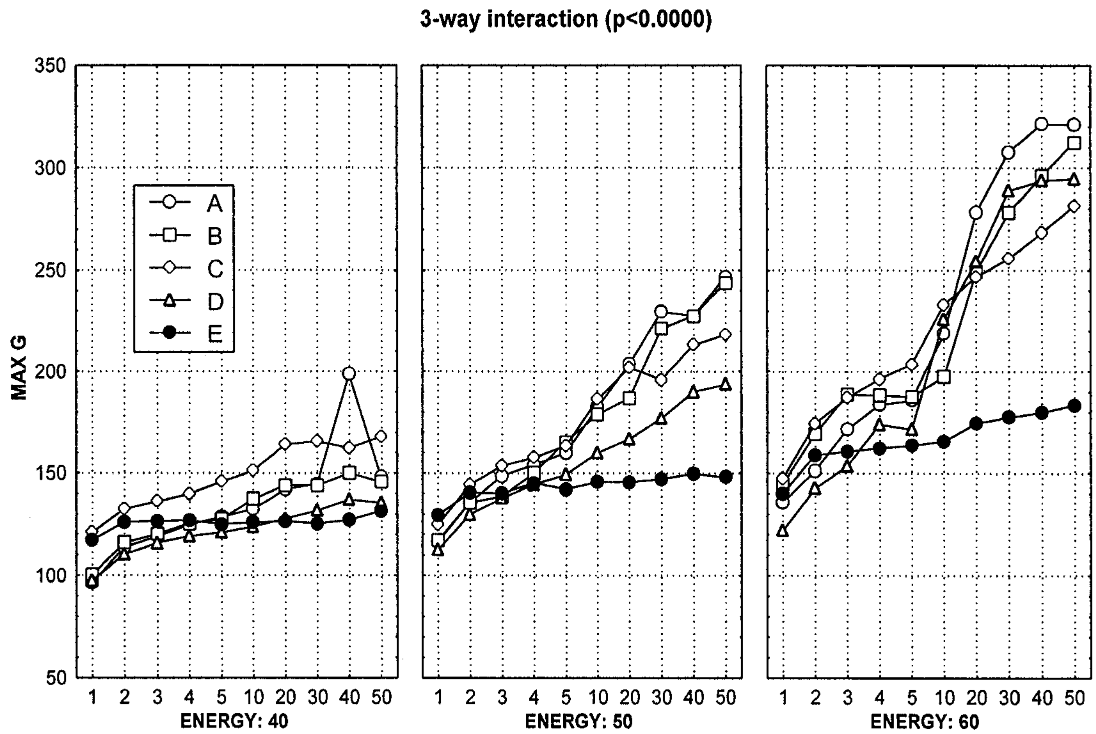


Figure 15: Comparison of peak G between the helmet models, energy levels, and the trials (see the text for details).

Figure 15 shows the three way interaction for the helmet models, energy levels and the trials; all the mean for all the helmet models were significantly different ($p<0.001$) for all the energy levels, except for the model E; for this model, there was only significant difference ($p<0.001$) between 40 and 60 J. At the 40 J, there were no statistical differences between the models, except between the models D and C. At the 50 J, the models A, B, and C were significantly different with E. At the 60 J, all the models were significantly different ($p<0.001$) with the model E. For all the models, there were significant differences ($p<0.001$) between all trials with the first trial, except the model E; only trials 20, 30, 40 and 50 were significantly different ($p<0.001$) from the first trial. There were significant differences ($p<0.001$) between the trials; however, there was no significant difference between each trial with its previous and the next trial. There was no significant difference between the trials for the model E. This model has significant difference ($p<0.001$) with the other models after the 5th impact.

4.4. Repeated impacts at 40 J

In this step, the analysis was conducted for all the helmet models, all the sites, and the first 10 trials at the energy level of 40 J. This was to evaluate the peak acceleration attenuation properties of the helmets and their sites at 40 J (table 7-11 and 14). Table 14 shows significant main effects and interactions for all the factors ($p< 0.05$).

Table 14: ANOVA values for peak Gs including all the variables at 40 J (up to 10 trials)

1-MODEL, 2-SITE, 3-TRIAL						
	Df Effect	MS Effect	df Error	MS Error	F	p-level
1	4	20464.96	40	588.46	34.78	0.00
2	3	144505.73	40	588.46	245.56	0.00
3	5	20878.44	200	135.61	153.96	0.00
12	12	15205.31	40	588.46	25.84	0.00
13	20	264.71	200	135.61	1.95	0.01
23	15	5039.86	200	135.61	37.16	0.00
123	60	772.28	200	135.61	5.69	0.00

From the different post hoc tests (Tukey test) significant differences were found. Comparison between the models showed that there were significant differences ($p<0.001$) between all the models, except there were no significant differences between the models C and E.

Comparison between the sites showed significant differences ($p<0.001$) between all the sites with the side impacts at the 40 J; moreover, the crown and rear sites were significantly different. There were no significant differences between the sites for the model A. For the crown impacts, the model C was significantly different ($p<0.001$) with the other models. There were no significant differences between the front and the rear sites with the other sites. The side impacts had significant difference ($p<0.001$), except between the models C with D, D with E, and E with D.

Figure 16 shows that all the trials were significantly different ($p<0.001$) from the 1st trial for all the helmet models; in addition, the 10th trials also had significant differences ($p<0.001$) with the 2nd trials. Between the models, the following significant differences ($p<0.001$) were existed: the model A was significantly different ($p<0.001$) from the models C and E, and after 2nd trial from the model D; it had no significant difference from the model B; there were no significant differences between B and D; the model B was significantly different from C and E after the 3rd trial.

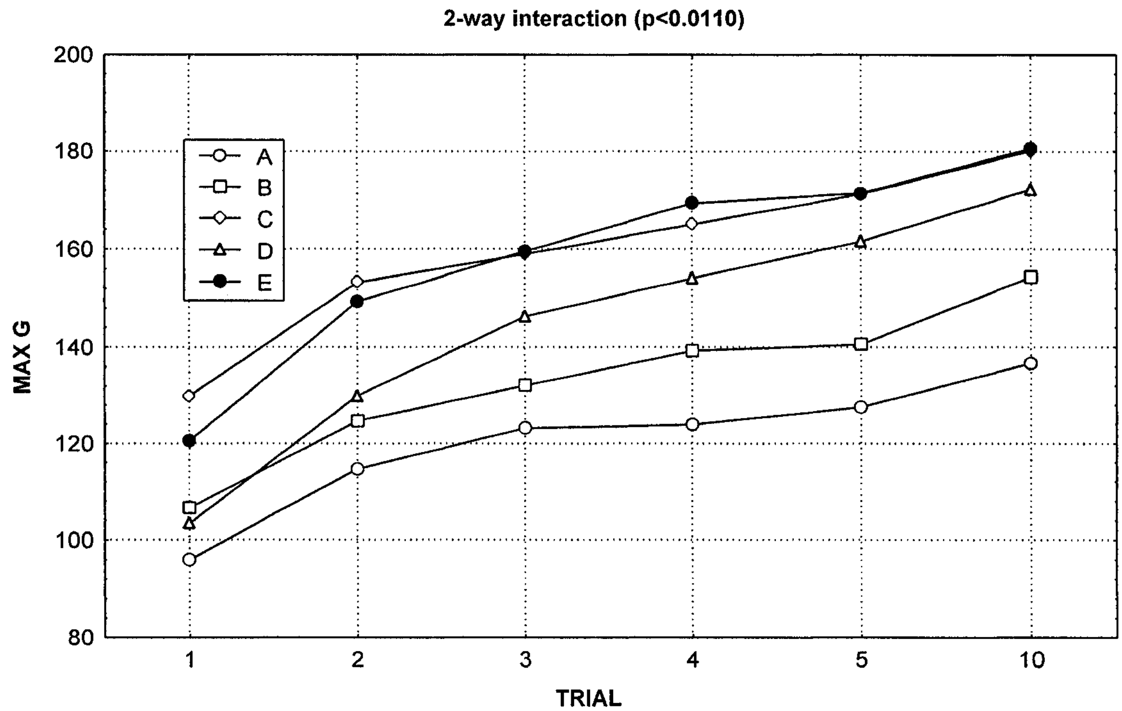


Figure 16: Comparison of peak G between the trials and the helmet models at 40 J (see the text for more details).

Figure 17 shows that for all the sites, there were significant differences ($p<0.001$) between all the trials with the first trial, except for the crown between 2nd and 1st trial; for the rear impacts, there was statistical difference only between 10th trial with the 1st trial. The side impacts were significantly different ($p<0.001$) with all the other sites at all the trials, except at the first trial with the crown impact. There were no significant differences between the crown, rear and front for all the trials, except between the crown and rear sites at 5th impact.

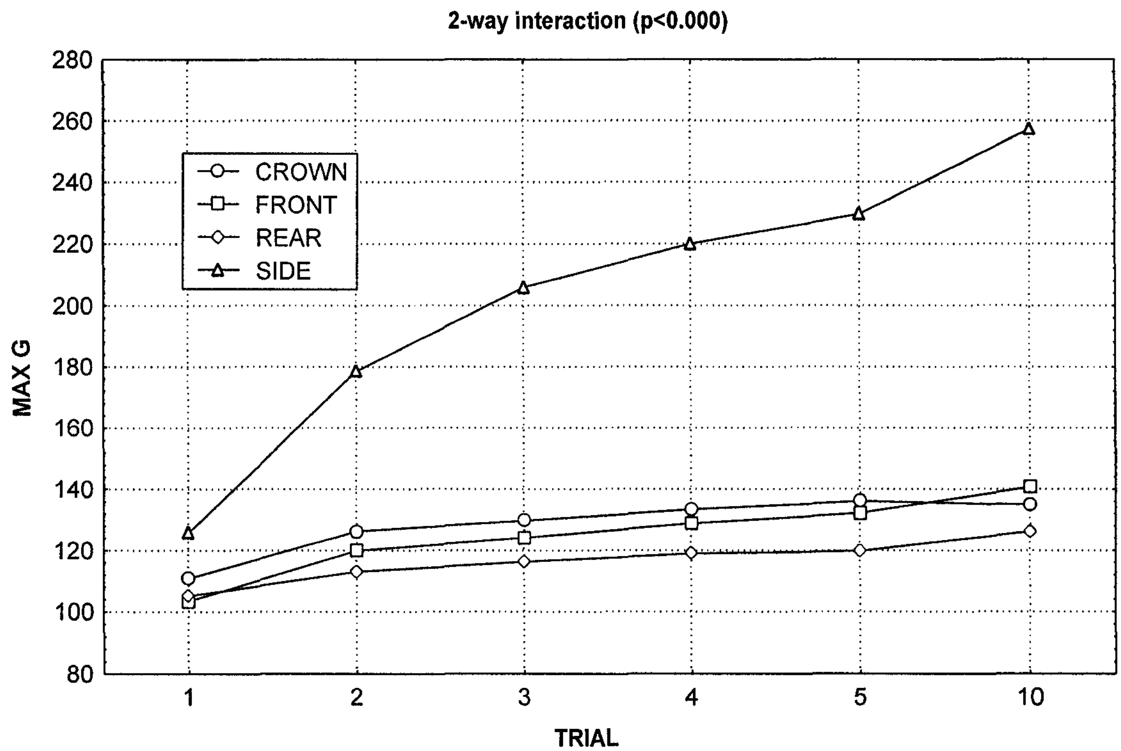


Figure 17: Comparison of peak G between the sites and the trials at the 40 J (see the text for details).

Figure 18 shows that by increasing the trials the side impacts were getting more substantial for all the models except the model A; however, the side impacts for the model A were significantly different ($p<0.01$) from the other models at all the trials except the 1st trial. This figure represents all the significant differences from the figures 16 and 17.

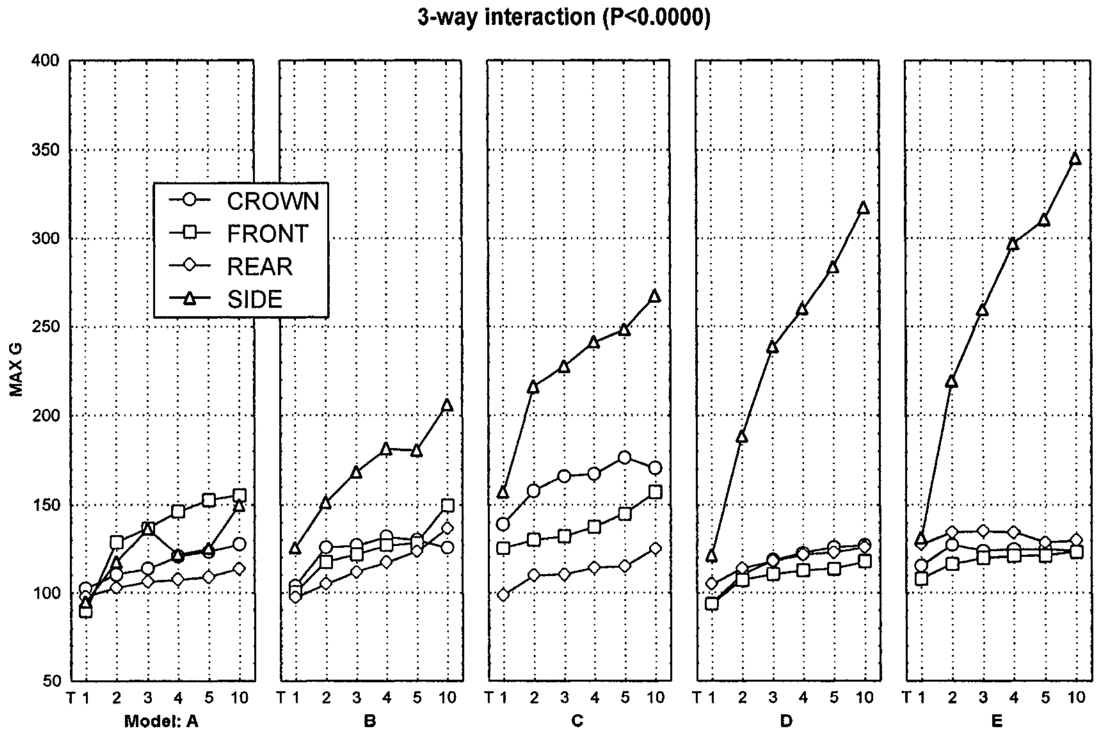


Figure 18: Comparison of peak G between the sites and helmet models at the firs 10 trials for the 40 J (see the text for detail).

4.5. Repeated impacts at 50 J

In this step, the analysis was conducted for all the helmet models, all the sites, and the first 5 trials at the energy level of 50 J. This was to evaluate the peak acceleration attenuation properties of the helmets and their sites at 50 J (table 7-11 and 15). Table 15 shows significant main effects and interactions for all the factors ($p < 0.05$).

From the different post hoc tests (Tukey test) significant differences were found; there were significant difference ($p<0.001$) between the model D and the models A, B, and C. There was significant difference ($p<0.001$) between the models A and E. Moreover, the side impacts were significantly different ($p<0.001$) from the other sites; there was no significant difference between the other sites. The side impacts were significantly different ($p<0.001$) from the other 3 sites for all the models; except, there was no significant difference between the side of model A with the front of the model A, rear of the model B, and crown and front of the model C. There was no significant difference between the sides of the models A, B and C; the sides for the models D and E were significantly different from all the models.

Table 15: ANOVA values for peak Gs including all the variables at 50 J (up to 5 trials)

1-MODEL, 2-SITE, 3-TRIAL						
	Df	MS	Df	MS		
	Effect	Effect	Error	Error	F	p-level
1	4	9713.87	40	1216.98	7.98	0.00
2	3	245501.50	40	1216.98	201.73	0.00
3	4	39992.95	160	145.84	274.23	0.00
12	12	18404.37	40	1216.98	15.12	0.00
13	16	775.20	160	145.84	5.32	0.00
23	12	8915.59	160	145.84	61.13	0.00
123	48	1105.73	160	145.84	7.58	0.00

Figure 19 shows that all the trials were significantly different ($p<0.001$) from the 1st trial for all the helmet models. The differences between the trials mostly were significant; the non-significant relations were placed between the trials with their closest trials. The model D was significantly different ($p<0.001$) with all the models; the models A and B were significantly different ($p<0.001$) from the model D after 2nd trial, the model C after 3rd trial, and the model E after 4th trial.

Figure 20 shows that the side impacts were significantly different ($p<0.001$) from the other sites at all the trials. There were significant differences ($p<0.001$) between all the trials with the 1st trial for all the sites; except for the rear impact, there was no significant difference between 1st and 2nd trial. There was no significant differences between the crown, front, and rear sites in all the trials.

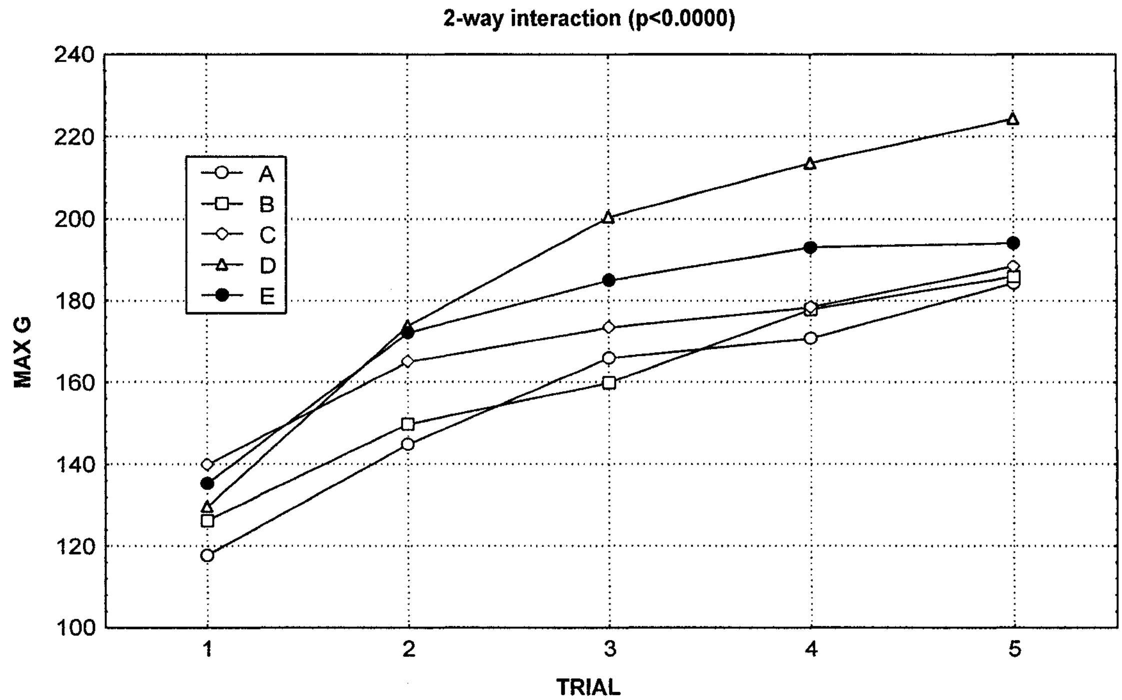


Figure 19: Comparison of peak G between the helmet models and the trials at 50 J (see the text for details).

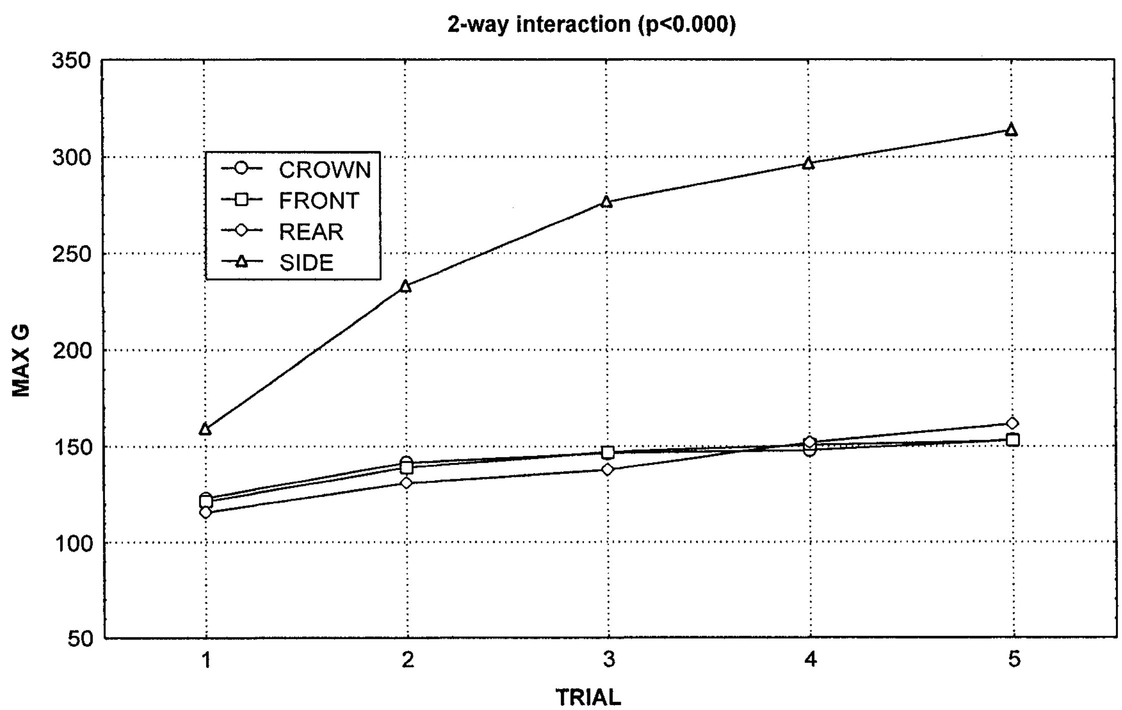


Figure 20: Comparison of peak G between the sites and the trials at 50 J (see the text for details).

Figure 21 shows that by increasing the trials the side impacts are getting more substantial for all the models; for the model A, there is no significant differences between the sites for the first two trials. This figure represents all the significant differences from the figures 19 and 20.

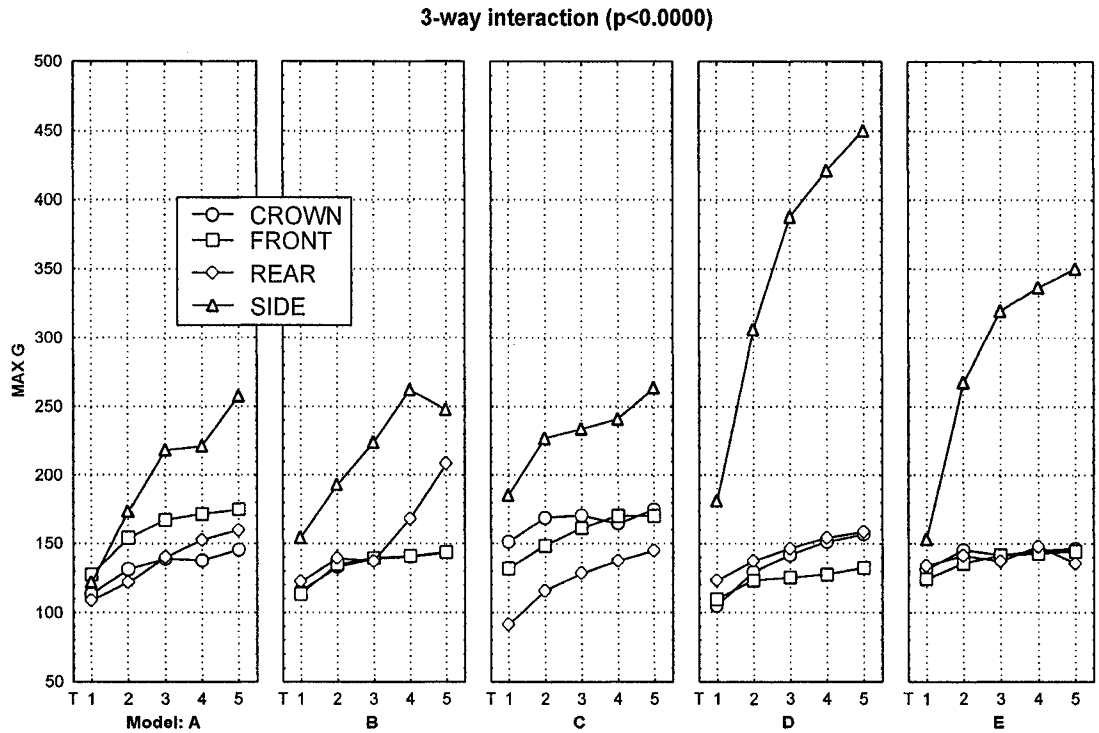


Figure 21: Comparison of peak G between the sites, helmet models, and the first 5 trials at 50 J (see the text for details).

4.6. Repeated impacts at 60 J

In this step, the analysis was conducted for all the helmet models, all the sites, and the first 3 trials at the energy level of 60 J. This was to evaluate the peak acceleration attenuation properties of the helmets and their sites at 60 J (tables 7-11 and 16). Table 16 shows significant main effects and interactions for all the factors ($p < 0.05$).

Table 16: ANOVA values for peak Gs including all the variables at 60 J (up to 3 trials)

1-MODEL, 2-SITE, 3-TRIAL						
	df Effect	MS Effect	Df Error	MS Error	F	p-level
1	4	12486.69	40	617.33	20.23	0.00
2	3	189513.94	40	617.33	306.99	0.00
3	2	58127.88	80	212.23	273.90	0.00
12	12	25744.39	40	617.33	41.70	0.00
13	8	981.30	80	212.23	4.62	0.00
23	6	11787.22	80	212.23	55.54	0.00
123	24	1685.32	80	212.23	7.94	0.00

From the different post hoc tests (Tukey test) significant differences were found; Comparison between the models at 60 J showed that the model A was significantly different ($p<0.001$) from the other models. The other models had significant differences ($p<0.001$); there was no significant difference between the model C and B, C and E, D and E.

Comparison between the sites, and the trials and the sites at 60 J showed that the side impacts were significantly different ($p<0.001$) from the other sites; there was no significant difference between the crown, front, and rear sites. There were significant differences ($p<0.001$) for each site from the 1st trial to the 3rd trial. There were no significant differences between 2nd and 3rd trials for the crown and front impacts.

Comparison between the helmet models and the sites at 60 J showed that there were no differences between the sites for the model A. For the model B, the crown and front sites were significantly different ($p<0.001$) from the side; moreover, the crown and the rear were significantly different ($p<0.001$). The side impacts were significantly different from the other sites for the models C, D, and E; there was no significant difference between the other sites, except for the model C, between the crown and rear impacts.

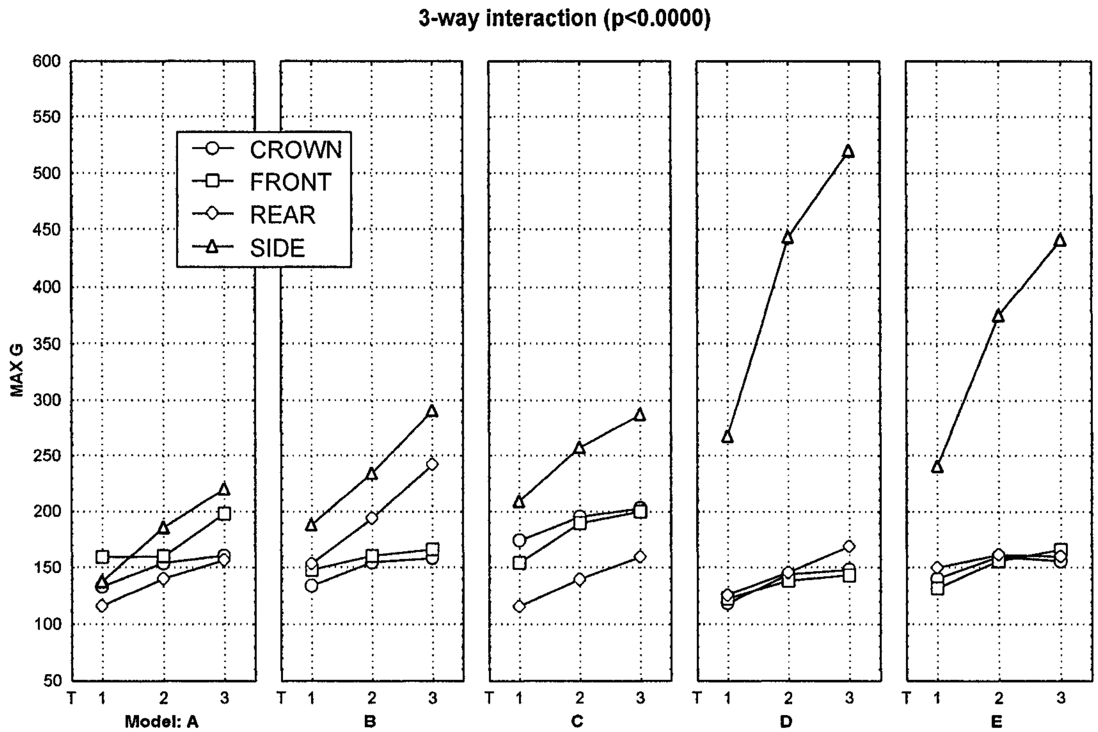


Figure 22: Comparison between the sites, helmet models, and the firs 3 trials at 60 J (see the text for details).

Figure 22 shows that by increasing the trials the side impacts were getting more substantial for all the models. For the model A, the only significant difference ($p<0.001$) between the sites, was between the 1st and the 3rd side impact.

Comparison between the helmet models and the trials showed that there were significant differences ($p<0.001$) for each site from the 1st trial to the 3rd trial. There were no significant differences between 2nd and 3rd trials for the models B and E. The model A was significantly different ($p<0.001$) from the other models at all the trials, except from the model B at the 1st trial. The models B, C, and D were significantly different ($p<0.001$) at the 2nd and 3rd trial. The models B and E were significantly different ($p<0.001$) at the 2nd trial.

Comparison between the sites and the trials showed that the side impacts were significantly different ($p<0.001$) from the other sites at all the trials. There was no significance between the crown, front, and rear sites for all the trials. There were significant differences between all the trials for the side and rear impacts. For the crown and front impacts, there were significant differences ($p<0.001$) between the 1st and 3rd trials.

4.7. Peak G prediction function: regression analysis for multiple impacts

Peak G data were graphed with respect to impact number and trend lines with regression functions generated using the Microsoft® Excel (2002). The peak G values for the three trials for the three samples were averaged and graphed for each model and each site. Figures 23-27 show the average peak acceleration for three trials at three energy levels for each site of each helmet model. The best fit logarithmic trendlines were based on regression anlaysis. High R-squared values (i.e. $R^2 > 0.7$ or 70% of the variance) showed good prediction of helmet-site response with respect to multiple impacts (table 17); however, in almost one fifth of the plots the R-squared values were very small (i.e.$R^2 < 0.7$).

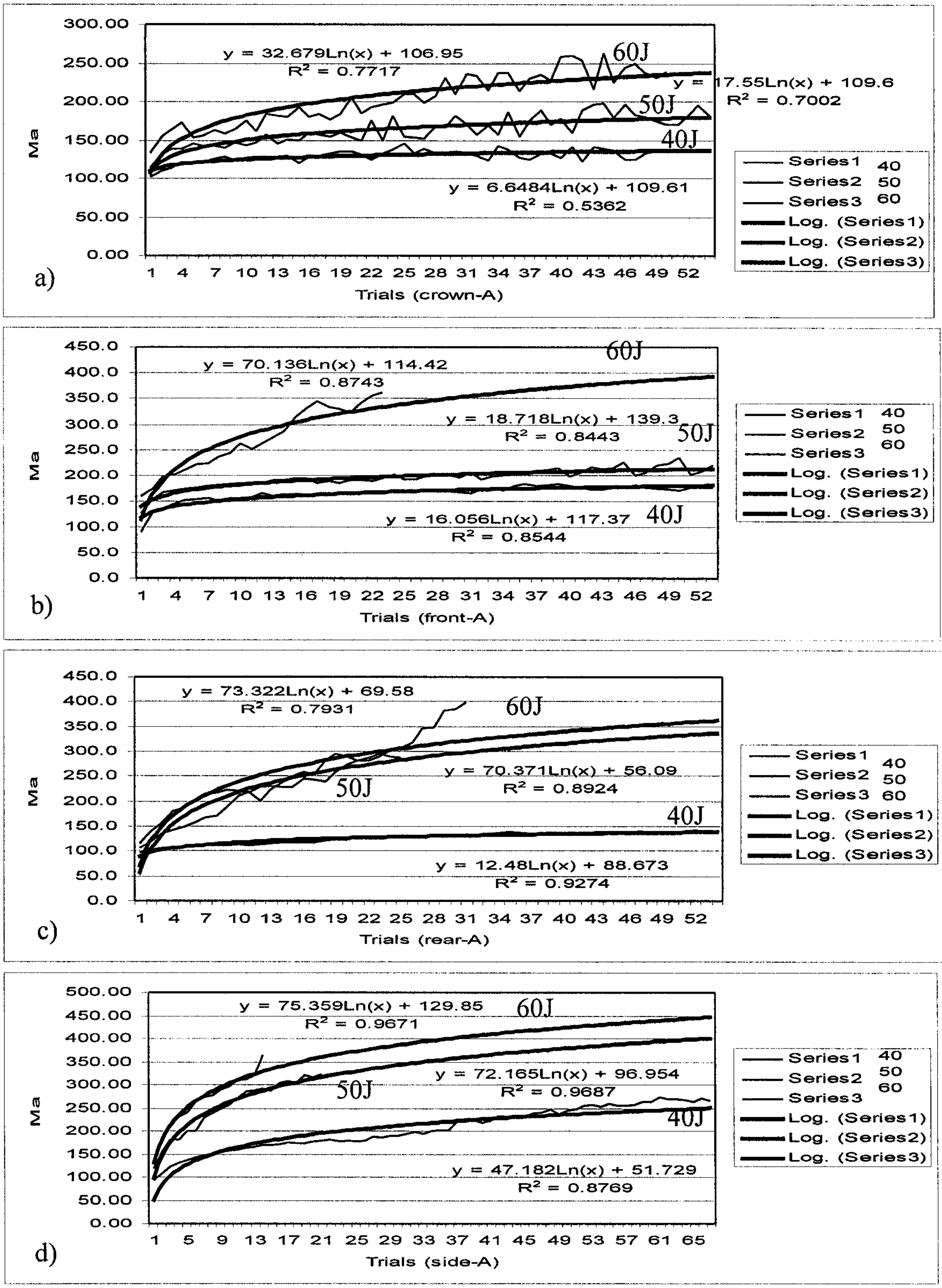


Figure 23: Average peak values for three samples of the helmet model A at 40, 50, and 60 J. Helmet sites include: crown (a), front (b) rear (c) and side (d). Y axis shows Max G.

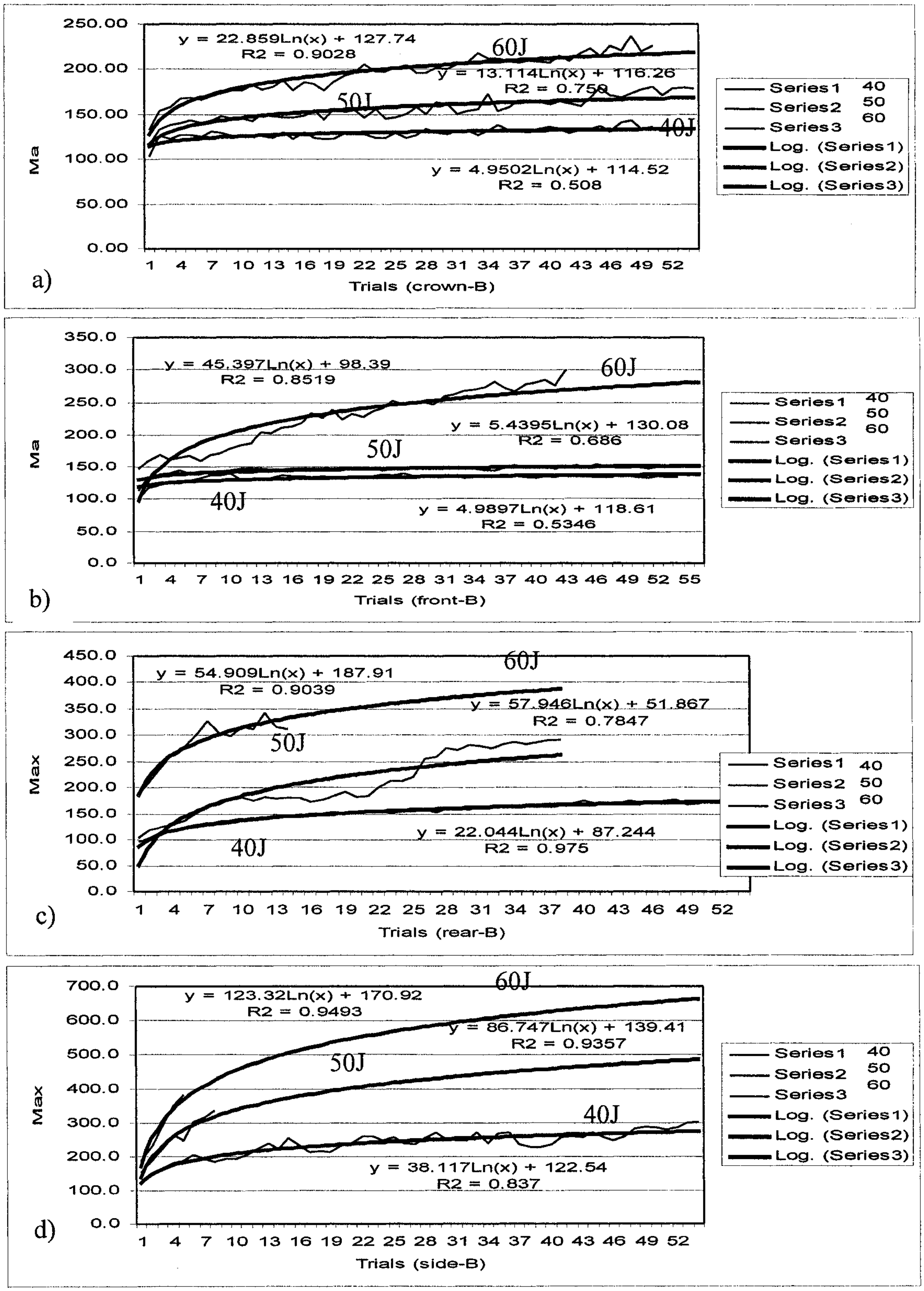


Figure 24: Average peak values for the three samples of the helmet model B at 40, 50, and 60 J. Helmet sites include: crown (a), front (b) rear (c) and side (d). Y axis shows Max G.

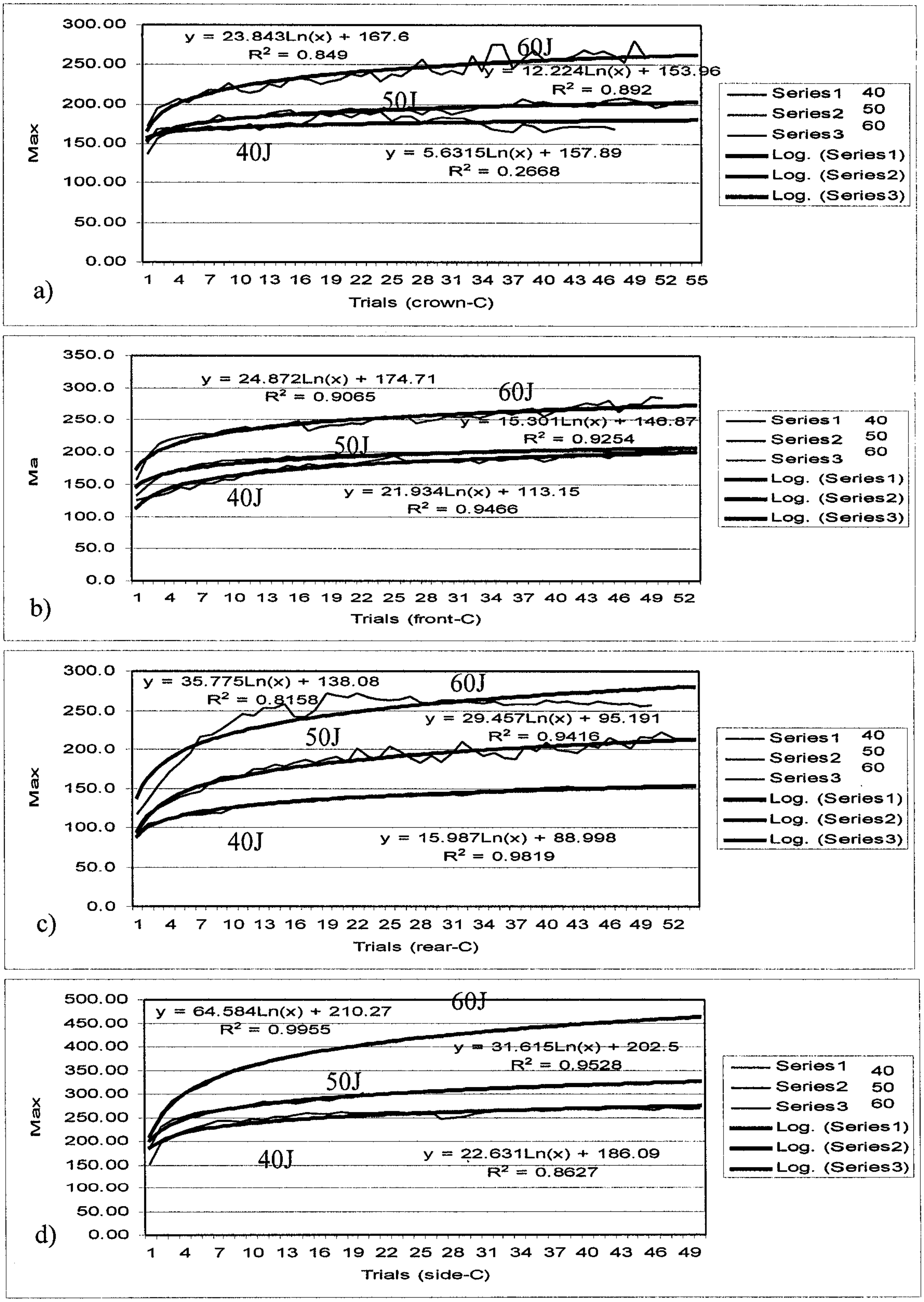


Figure 25:Average peak values for the three samples of the helmet model C at 40, 50, and 60 J. Helmet sites include: crown (a), front (b) rear (c) and side (d). Y axis shows Max G.

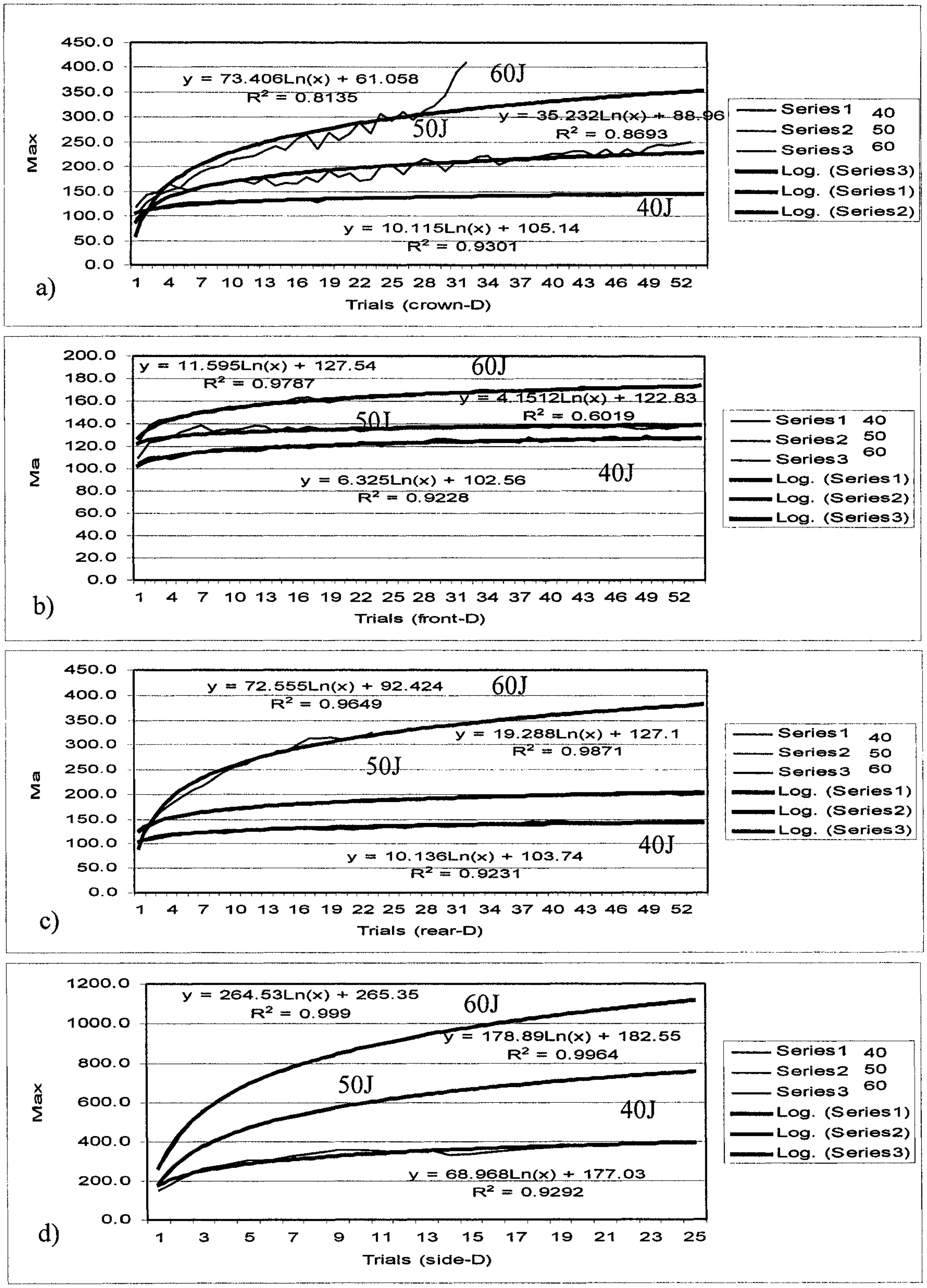


Figure 26: Average peak values for the three samples of the helmet model D at 40, 50, and 60 J. Helmet sites include: crown (a), front (b) rear (c) and side (d). Y axis shows Max G.

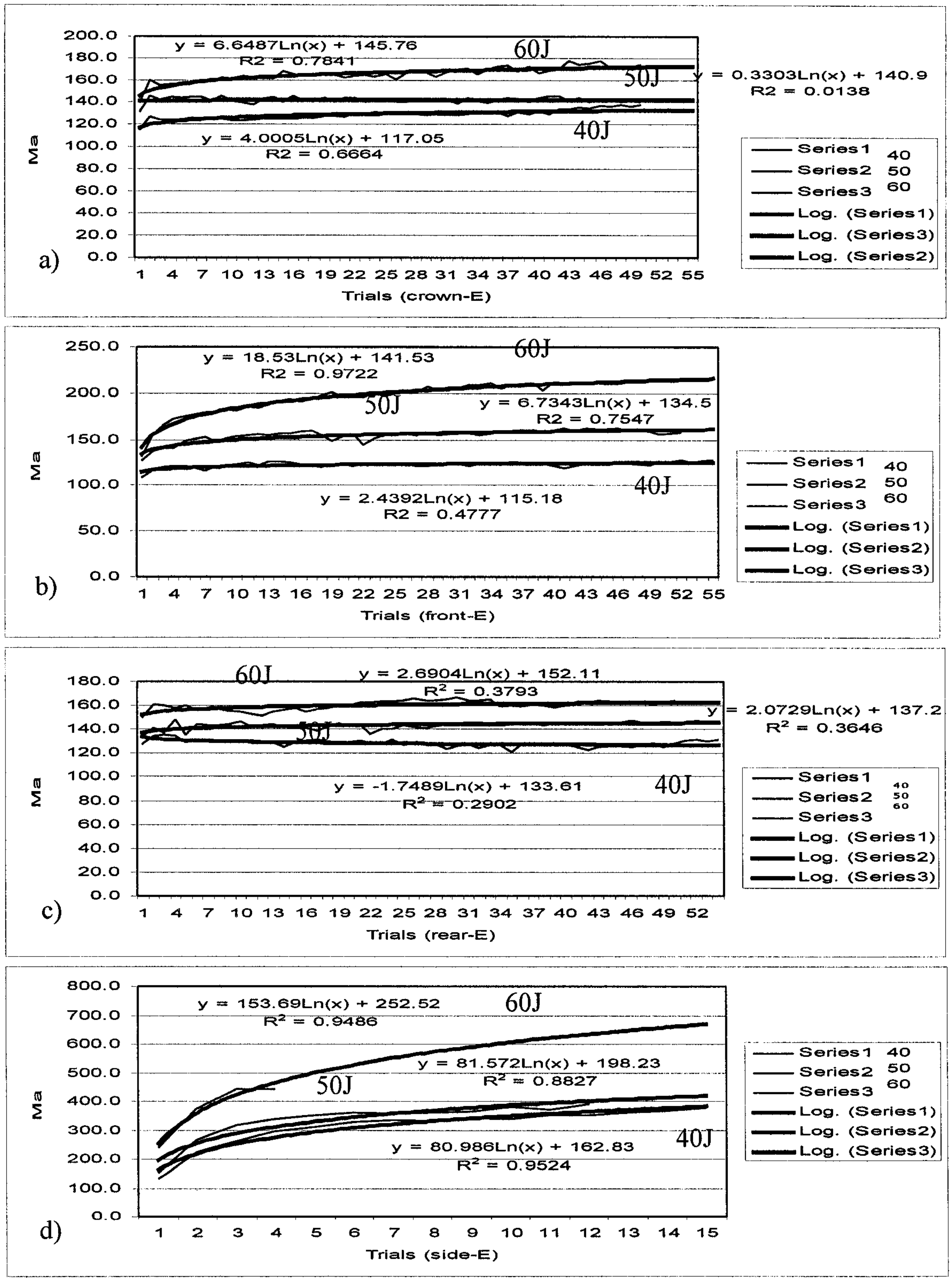


Figure 27: Average peak values for the three samples of the helmet model E at 40, 50, and 60 J. Helmet sites include: crown (a), front (b) rear (c) and side (d). Y axis shows Max G.

Table 17: Equations for the regression lines from the figures 23-27.
The logarithmic trendlines were the best fit for the data set.

SITE	ENERGY	MODEL				
		A	B	C	D	E
Front	40	y = 16.056Ln(x) + 117.37 R^2 = 0.8544	y = 4.9897Ln(x) + 118.61 **R2 = 0.5346**	y = 21.934Ln(x) + 113.15 R2 = 0.9466	y = 6.325Ln(x) + 102.56 R2 = 0.9228	y = 2.4392Ln(x) + 115.18 **R2 = 0.4777**
	50	y = 18.718Ln(x) + 139.3 R2 = 0.8443	y = 5.4395Ln(x) + 130.08 **R2 = 0.6869**	y = 15.301Ln(x) + 146.87 R2 = 0.9254	y = 4.1512Ln(x) + 122.83 **R2 = 0.6019**	y = 6.7343Ln(x) + 134.5 R2 = 0.7547
	60	y = 70.136Ln(x) + 114.42 R2 = 0.8743	y = 45.397Ln(x) + 98.39 R2 = 0.8519	y = 24.872Ln(x) + 174.71 R2 = 0.9065	y = 11.595Ln(x) + 127.54 R2 = 0.9787	y = 18.53Ln(x) + 141.53 R2 = 0.9722
Rear	40	y = 12.48Ln(x) + 88.673 R2 = 0.9274	y = 22.044Ln(x) + 87.244 R2 = 0.9758	y = 15.987Ln(x) + 88.998 R2 = 0.9819	y = 10.136Ln(x) + 103.74 R2 = 0.9231	y = -1.7489Ln(x) + 133.61 **R2 = 0.2902**
	50	y = 70.371Ln(x) + 56.09 R2 = 0.8924	y = 57.946Ln(x) + 51.867 R2 = 0.7847	y = 29.457Ln(x) + 95.191 R2 = 0.9416	y = 19.288Ln(x) + 127.1 R2 = 0.9871	y = 2.0729Ln(x) + 137.2 **R2 = 0.3646**
	60	y = 73.322Ln(x) + 69.58 R2 = 0.7931	y = 54.909Ln(x) + 187.91 R2 = 0.9039	y = 35.775Ln(x) + 138.08 R2 = 0.8158	y = 72.555Ln(x) + 92.424 R2 = 0.9649	y = 2.6904Ln(x) + 152.11 **R2 = 0.3793**
Side	40	y = 47.182Ln(x) + 51.729 R2 = 0.8769	y = 38.117Ln(x) + 122.54 R2 = 0.8375	y = 22.631Ln(x) + 186.09 R2 = 0.8627	y = 68.968Ln(x) + 177.03 R2 = 0.9292	y = 80.986Ln(x) + 162.83 R2 = 0.9524
	50	y = 72.165Ln(x) + 96.954 R2 = 0.9687	y = 86.747Ln(x) + 139.41 R2 = 0.9357	y = 31.615Ln(x) + 202.5 R2 = 0.9528	y = 178.89Ln(x) + 182.55 R2 = 0.9964	y = 81.572Ln(x) + 198.23 R2 = 0.8827
	60	y = 75.359Ln(x) + 129.85 R2 = 0.9671	y = 123.32Ln(x) + 170.92 R2 = 0.9493	y = 64.584Ln(x) + 210.27 R2 = 0.9955	y = 264.53Ln(x) + 265.35 R2 = 0.999	y = 153.69Ln(x) + 252.52 R2 = 0.9486
Crown	40	y = 6.6484Ln(x) + 109.61 **R2 = 0.5362**	y = 4.9502Ln(x) + 114.52 **R2 = 0.5086**	y = 5.6315Ln(x) + 157.89 **R2 = 0.2668**	y = 10.115Ln(x) + 105.14 R2 = 0.9301	y = 4.0005Ln(x) + 117.05 **R2 = 0.6664**
	50	y = 17.55Ln(x) + 109.6 R2 = 0.7002	y = 13.114Ln(x) + 116.26 R2 = 0.759	y = 12.224Ln(x) + 153.96 R2 = 0.892	y = 35.232Ln(x) + 88.96 R2 = 0.8693	y = 0.3303Ln(x) + 140.9 **R2 = 0.0138**
	60	y = 32.679Ln(x) + 106.95 R2 = 0.7717	y = 22.859Ln(x) + 127.74 R2 = 0.9028	y = 23.843Ln(x) + 167.6 R2 = 0.849	y = 73.406Ln(x) + 61.058 R2 = 0.8135	y = 6.6487Ln(x) + 145.76 R2 = 0.7841

5. Discussion

From the above results, it is apparent that all main factors (i.e. helmet model, impact energy, impact repetition, impact site) had a significant influence on the mechanical response of impact attenuation. Furthermore, significant interactions between these factors were also revealed. Consequently, to define the durability of a helmet requires consideration of all these factor dimensions. The implication of each factor will be discussed in turn.

5.1. General Findings

As hypothesized, the ability of ice hockey helmets to attenuate impact energies decreased after impact energies increased and / or the number of impacts increased. The findings indicate that cumulative, permanent changes to the helmet, in particular the foam liners, must have occurred due to tearing and / or crushing, which in turn, decreased the effectiveness of helmets to dissipate focal impact energies. Though the impact protocol followed represents an extreme of physical usage, it does clearly indicate the outer range of durability of these protective head gear. In most instances, the helmets were exceptionally robust, demonstrating minimal deleterious impact attenuation changes at 40J for the sites of the crown, front and rear of the helmet. In contrast, alarmingly poor durability was observed at the side impact site; in particular, two helmet models were below the accepted threshold limit after only four or five repeated impacts at 40J. Therefore, these results underscore that ice hockey helmets have a delimited durability. Though, they perform well within the given standard, the evidence that a gradual decrease in the liner's performance beyond three impacts is apparent. As a consequence, a stated time period for normal use of ice hockey helmets should be considered.

The findings of this study showed that different helmet models have variable durability and as well as behave differently for multiple impacts. Different padding foams used in inner liners, their shape, and thickness, as well as the geometry of the outer shells can be accountable for their behavior. Although all the helmets meet the CSA standards (CAN/CSA-Z262.1-M90), attenuation properties were found to be substantially reduced beyond three repeated impacts and above 40 J of impact energy. At 40 J of impact energy almost all the helmets passed 50 repeated impacts at all the sites, except the models D and E at the side impacts (table 5). In particular, these two models exceeded the peak G threshold after only 4 impacts at the side, where as the other helmets could sustain up to 45 impacts. (Note that the side site tests were confirmed by repeating the tests for each helmet on the left side) The liner used for the models C, D, and E is EPP foam; thus, the liner alone cannot be the reason to be different impact performance at the side. Construction differences are suspect; for instance, the small padding area at the side, and a 2.5 cm gap between the front and the rear padding could be the reason for the poor impact performance at the side for the D and E models (figure 28). Yet, these two helmet models showed excellent energy attenuation properties at the other sites even at higher energy levels (table 6). Further detailed analysis of side impact behaviour is warranted.

a

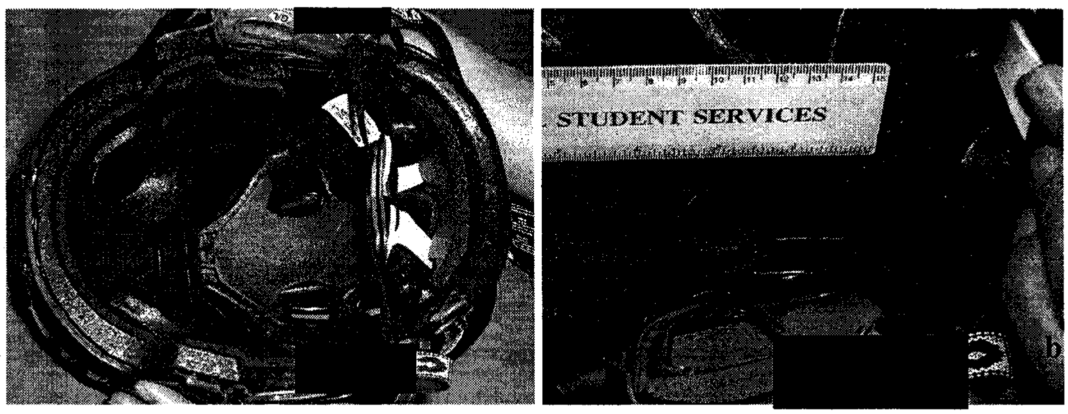


Figure 28: A 2.5 cm gap between the front and rear padding of the helmet:
a) interior view of helmet, b) close up of side foam geometry.

Tables 5 and 6 showed that as the impacts become more severe, the helmets absorbed less energy and therefore transferred more energy to the head. Thus, the headform accelerations increased beyond the threshold safety limit with a fewer number of impacts. One exception was observed at the rear impacts for the A model: although the acceleration range increased from 50 to 60 J of energy (107-289 vs. 116-399), the helmets exceeded the peak G threshold earlier at 50 J than 60 J (an average of 18±4 vs. 21±7 impacts, respectively). These latter differences were not statistically significant ($p<0.001$). This outcome may have resulted from subtle helmet positioning and orientation differences during testing. For instance, if the primary impact location was 5 to 10 mm lower than the rear site due to shell geometry projections (as in the case of A and B models), the headform acceleration could vary substantially. Similar observations existed with the front impacts. If the impact was only 5 to 10 mm offset, the number of impacts until peak G threshold could change by 2 to 5 trials. These differences at the rear and front sites could be a point of contention in the standards, because it is possible to manipulate the peak G measures by minor displacement of the impact point.

5.2. Helmet Model

The statistical analyses showed significant differences ($p<0.00$) between the helmet models. Studies by Pearsall et al. (2000) and Bishop and Briard (1984) reported similar findings. In general, the A and B models had the best overall performance (figure 6) when the statistical analyses were conducted for all combinations of helmet, energy, and site, within the first three trials. There were no statistical differences between the other models at this step (section 4.2.). However, by eliminating the side impact site from the analysis and including all 50 trials, the D and E models demonstrated the best performance. There were no statistical differences between the other models at this step (section 4.3.). Table 6 shows that the peak Gs were lower for the D and E models at the crown, front, and rear impacts. The A model (figure 6) showed the highest energy attenuation properties at all energy levels from 40, 50 to 60 J, followed by the model B. Differences in energy attenuation are due to the different liner properties (e.g. thickness, density, shape) and / or geometry of the helmets.

One problem of note with repeated impacts was the unanticipated movement of helmet around the headform that in turn changes the impact location (up to 10 mm) between trials. The extent of helmet-to-headform conformity and slippage can influence the variability of impact results. For instance, it was difficult to secure some helmets to the headform to avoid slippage; in particular, helmet models A and B. Chin straps had reduced effect given that the headform lacks facial features to bind onto. The issue of helmet movement relative to the headform needs to be addressed in future studies. It should be noted that helmet slippage is not necessarily a undesirable feature, as it may aid in reduction of rotation angular accelerations as well as avoid skin lesions.

5.3. Site

Analyses showed that the side impacts were significantly different ($p<0.001$) from the other sites for all the helmet models, energy levels, and trials (figures 7, 8, 9, 11), with the exception of the model A at the energy of 40 J. The side of the helmets demonstrated less energy attenuation properties. Similar to previous studies (Bishop and Arnold, 1993; Bishop and Briand, 1984; Gilchrist and Mills, 1996), this study showed that the sides of the helmets were not as effective in eliminating energy of local impacts. This is disturbing since, the side of the head is often quoted as being the most vulnerable region of the skull to fracture due to thin bony area (Gurdjian, 1975). Hence efforts should be made to improve helmet design for the impact attenuation at this side site.

There were no significant differences between the crown, front, and rear sites; however, there were some significant differences ($p<0.001$) between the sites for different helmet models. Overall, the model A had best performance at the side; the model E had best performance at the crown and rear; and the model D had best performance at front. By increasing the impact energy and / or the number of trials, significant differences ($p<0.001$) occurred for each site (figures 8, 11, 14, 18, 21). The rear site showed less energy absorbing behavior compare with the crown and front (table 5).

5.4. Impact Energy

By increasing the impact energy, the effectiveness of the helmets was decreased (table 5). The helmets transferred more energy and absorbed less; thus, the average peak accelerations of headform were increased substantially (from 4% to 80% after first impact) for all the helmets (table 6, figures 8, and 10). In general, the helmets performed well at the 40 J (the energy level for the standards); however, at 50 and 60 J of impact energy, their ability to attenuate the impact force was reduced substantially, especially at the side. Peak Gs were increased tremendously from the 1st trial to the 2nd trial at the side impacts with 60 J of impact energy (table 6). Table 5 shows that even one impact was enough to surpass the peak G threshold level of 275 g. There were significant differences ($p<0.001$) between the helmets at different energy levels (figures 10 and 15).

At the 60 J of impact energy, two helmets (A and D) sustained broken shells (at the rear and side impacts), one helmet (E) exhibited cracked foam liner (rear impacts), and in one helmet (E) the foam liners were separated from the outer shell (front impact). At 50 J, one helmet (D) sustained a broken shell (crown impact) and in two helmets (C and A) the foam liners were separated from the outer shells (front and rear impacts). Separation of liners from the outer shells did not change the outcomes, because the liners were gripped on the headform. All three shell cracks occurred at the crown area during crown, rear, and side impacts; this is analogous to skull fractures that also do not necessarily occur at the point of impact due to inhomogeneous bone strength distribution of the skull (Gennarelli, 1991a; Gurdjian, 1975).

5.5. Trial

Repeated impacts had a significant influence on the energy absorbing behavior of helmets. All trials were significantly different ($p<0.001$) from the first trial for all the helmets at all the energy levels (figures 12, 15, 16, and 19). This is similar to the finding by Spyrou et al. (2000) where significant differences ($p<0.05$) were found between all three trials. Some of the observed variation in peak G measures were in part due to the difficulty of impacting the exact same point for repeated impacts. Rotation of helmet around the headform was the main reason for this variance. Though helmets were all claimed to be of the same size (i.e. large size with circumference of 55 mm), they fit differently on the standard headform; thus, some helmet could slip and rotate around the headform on impact. Though helmets were repositioned, the impact site may have varied by up to 10 mm.

Each impact can cause cumulative changes in the energy absorbing property of a helmet without leaving visual deformation or any detectable changes on it. From the total of 45 helmets, only seven helmets had shown major damages in the energy levels of 50 and 60 J; no damage was occurred at 40 J.

Figures 23-27 show the average peak acceleration for three trials at three energy levels for each site of each helmet model. All the helmets showed an increase in peak acceleration from one level of energy to the next level. The logarithmic trend lines were found the best fit for the data set i.e. peak G values increased with impact repetition but at an exponentially decreasing rate. Most of the trend lines showed good correlation between the trials (i.e. $R^2 > 0.7$) and the maximum acceleration values; though, in almost one fifth of the plots (table 17), the R-squared values are very small. Trendlines varied with helmet models and the samples due in part to slippage and rotation of the helmets around the headform; therefore, the acceleration of head form had irregular increases. For instance, at the front impact for the model B (figure 24), increase in acceleration at 40 J was small compare with 60 J. With these regression functions, a profile of helmet durability is available to aid manufacturers to predict the multidimensional mechanical behaviour of ice hockey helmets. As well, it may be used by safety standard committees to guide future policy and standard development.

5.6. Liner

The main liners of the helmets in this study were Expanded Polypropylene (EPP) and Vinyl Nitrile (VN). Other comfort padding (poly vinyl chloride, PVC) in some models (C, D, and E) was included to eliminate pressure points and grip the head. Visula observation during testing noted that helmets with EPP liners (i.e. the model D) had higher elastic rebound than the helmets with VN liner. This could be interpreted that the VN liner has better energy absorbing properties. However, Spyrou et al. (2000) reported that the EPP liner could withstand higher impact energy than the VN liner.

Although the helmets with VN liner demonstrated better performance than the helmets with EPP liner, it is erroneous to attribute the results solely to these two types of liner. Three models had EPP liners. One of them had moderate performance (model C) and two of them had excellent performance at the crown, front, and rear sites (models D and E). The model C showed a steady increase in the head form acceleration from consecutive trials at the side impacts, but the models D and E showed a more rapid increases. Therefore, superiority of one type of liner to the other can not be identified from the performance of the helmets alone. The liners have to be tested separately with their test standards, rather than helmet testing. The material property of the outer shell and the thickness and shape of the inner liner can alter the results (Spyrou et al, 2000).

The importance of geometry of the liner was evidenced at the front impacts with 60 J at the model A and B. The model A reached peak G threshold after 13 impacts, while model B sustained 42 impacts. Although the inner liner for the both models is the same material (VN liner) and same thickness (16.5 mm), the covered area with the liner was different. For the model A, the intersection of front and rear pads were underneath the area of frontal impacts (Figure 29). Thus the front padding might not absorb the impact energy as well. The front padding for the model B covers all front and a part of side area.

One interesting finding concerned the deterioration pattern at the model C helmet. Table 5 showed that this model had a different deterioration pattern than the other models. The values for the Severity Index (SI) were reached threshold magnitude (1500) earlier than the threshold acceleration value (275 g). Note that the SI calculates the cumulative impulse according to the equation 1 (SI= $\int a^{2.5}(t)dt$).

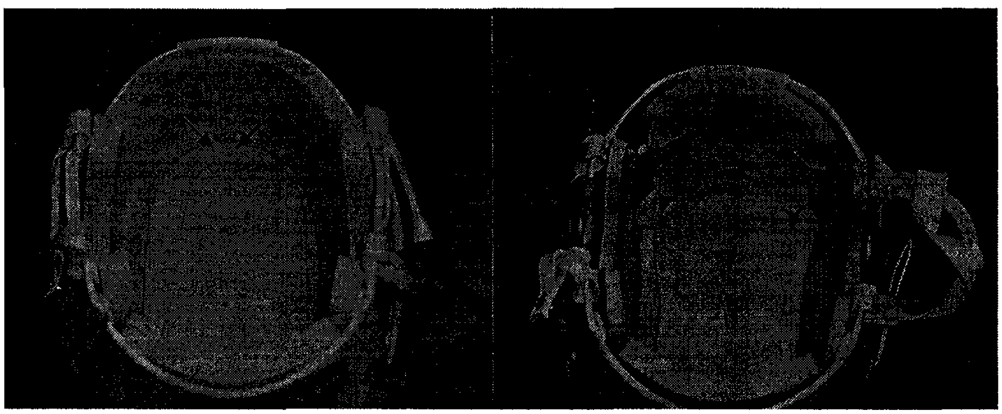
a  b

Figure 29: Comparison of the frontal padding between two models:
a) two crown and frontal pads are separated from each other at the area of front impacts. Arrows show the separation line. b) The frontal pad covers all front and a part of the crown area.

The main liners (i.e. EPP and VN) did not show any visual changes after multiple impacts; only in one helmet the foam was cracked at rear impacts with 60 J and in 3 helmets the foam were separated from the outer shell (front and rear impacts at 50 and 60 J). The comfort padding helped to have better fitting by gripping the head form during repeated impacts; thus, less helmet rotation occurred around the head form.

6. Conclusion

The results from this series of tests show a gradual decrease in the performance of the helmets with repeated impacts. Moreover, the findings of this study showed that different helmet models have variable durability and behave differently with multiple impacts. Different padding foams used in inner liners, their shape, and thickness, as well as the geometry of the outer shells can be accountable for their behavior. The results from 50 repeated impacts can help us to predict normal and extreme impact behaviour of helmets more extensively. This study suggests that manufacturers and / or safety standard organizations should consider stating a valid life-time usage for helmets. However, the implications of stating a product viability may have economic and social ramifications that need to be considered. Safety standard committees, manufactures and national ice hockey associations need to address these issues. In addition, further study is needed to determine the typical mechanical stability of helmets over a normal season-to-season use.

From these findings, a further proposal is that side impact attenuation performance of the helmets need to be improved. The results show that the sides of the helmets are not completely effective in attenuating the force of the impacts after three impacts at the temporoparietal region of the head. To improve helmet durability, manufactures need to focus on helmet shell and liner changes in this region.

References

Advani SH, Ommaya AK, and Yang WJ (1982). "Head injury Mechanisms: Characterizations and Clinical Evaluation." Book Analytic In: "Human body dynamics: impact, occupational, and athletic aspect." By: Ghista DN. Clarendon press, Oxford.

Allsop DL (1993). "Skull and Facial Bone Trauma: Experimental Aspects." Book Analytic In: "Accidental injury: Biomechanics and Prevention." By: Nahum AM and Melvin JW Springer-Verlag, pp 247-266, New York Inc.

Allsop DL, Warner CY, Wille MG, Schneider DC, and Nahum AM (1988). "Facial Impact Response: a comparison of the Hybrid III Dummy and Human Cadavers. 32nd Stapp, No. 881719, October 17-19.

Azuelos YS, Pearsall DJ, Turcotte R., and Montgomery DL (in press). "A Review of Ice Hockey Injuries: Location, Diagnosis, Mechanism and Context." Book Analytic in: "Safety in Ice Hockey." Fourth Volume, ASTM STP 1341, By: Ashare AB and Pearsall DJ, American Society for testing and Materials, West Conshocken, PA, 2002.

Bandak F (1995). "On the Mechanics of Impact Neurotrauma: A Review and Critical Synthesis." Journal of Neurotrauma, Vol. 12, No. 4, Mary Ann Liebert Inc.

Baroff GS (1998). "Is heading a soccer ball injurious to brain function?" J Head Trauma Rebabil, 13(2): 45-52.

Bishop PJ (1976). "Head Protection in Sport with Particular Application to Ice Hockey." Ergonomics, Vol. 19, No. 4, pp. 451-464.

Bishop PJ (1993). "Protective Equipment: Biomechanical Evaluation." Book Analytic In: "Sport Injuries: Basic Principles of Prevention and Care." By: Renstrom PAFH, Black Well Scientific Publications.

Bishop PJ (2000). "A Comparison of the Epoxy Alloy and Magnesium Alloy." Book Analytic In: Safety in ice hockey, third volume, ASTMSTP 1341, By: Ashare AB, American Society for testing and materials.

Bishop PJ and Arnold J (1993). "The effectiveness of Hockey helmets in limiting localized loading on the head." Book Analytic In: "Safety in ice hockey." By: Castalid, Bishop, and Hoernr, pp 175-182, USA.

Bishop PJ and Briard BD (1984). " Impact performance of bicycle helmets." Canadian journal of applied sport sciences, (Windsor, Ont.), vol. 9, no. 2, pp. 94-101.

Bruno LA (1991). "Focal Intracranial Hematoma" Book Analytic In: 'Athletic Injuries to the Head, Neck, and Face." Second edition, By: Torg JS, pp 305-322, Mosby year book

Clarke KS (1991). "An Epidemiologic View." Book Analytic In: "Athletic Injuries to the Head, Neck, and Face." Second edition, By: Torg J. S., pp 15-27, Mosby year book.

Dick RW (1993). "Injuries in Collegiate Ice Hockey." Book Analytic In: "Safety in Ice Hockey." Second Volume, By: Castaldi, Bishop, and Hoerner, pp 21-30.

Dixon JL (1989). "The Canadian Standards Association and the Evolution of Head and Face Protection in Canadian Hockey." Book Analytic In: "Safety in ice hockey." By: Castalid and Hoernr, pp 207-219, USA.

Dixon JL and Brodie IKR (1993). "The New ISO Standards for Ice Hockey Helmets and Face Protectors: Moving toward International Standards Harmonization and Conformity Assessment." Book Analytic In: "Safety in ice hockey." By: Castalid, Bishop, and Hoernr, pp 175-182, USA.

Douglas LA (1993). "Skull and Facial Bone Trauma: Experimental Aspects." Book Analytic In: "Accidental injury: Biomechanics and Prevention." By: Nahum AM and Melvin JW. Springer-Verlag, pp 247-266, New York Inc.

Gadd CW (1966). "Use of a Weighted-Impulse Criterion for Estimating Injury Hazard." 10th Stapp Car Crash Conference, Nov. 8-9, pp 164-174.

Gennarelli TA (1991a). "Cerebral Concussion and Diffuse Brain Injury." Book Analytic In: "Athletic Injuries to the Head, Neck, and Face." Second edition, By: Torg JS, pp 270-282, Mosby year book.

Gennarelli TA (1991b). "Head Injury Mechanisms" Book Analytic In: "Athletic Injuries to the Head, Neck, and Face." Second edition, By: Torg JS, pp 232-240, Mosby year book.

Gennarelli TA, Thibault LE, Adams JH, Graham DI, Thompson CJ, Marcincin RP (1982). "Diffuse axonal injury and traumatic coma in the primate." Ann Neurol. 12(6):564-74.

Gilchrist A and Mills NJ (1996). "Protection of the side of the head." Accid. Anal. And prev., vol.27, no.4, 525-535.

Gurdjian ES (1975). "Impact Head Injury: Mechanistic, Clinical, and Preventive Correlations" Charles C. Thomas Publisher, USA.

Gurdjian ES and Gurdjian ES (1975). "Re-evaluation of the Biomechanics of Blunt Impact Injury of the Head." Surgery Gynecology and Obstetrics, 140(6): 845-850.

Gurdjian ES, Lissner HR, and Patric LM (1962). "Protection of the head and neck in sports", Journal of American Med. Assoc., 182: 509-512.

Gurdjian ES, Thomas LM, and Lissner HR (1965). "High-speed Cinematographic Study of the Position and Motion of the Head at Impact." Transaction of the American Neurological Association. 90: 158-61.

Halstead PD, Alexander CF, Cook EM, and Drew RC (2000). "Hockey Headgear and the Adequacy of Current Design and Standards." Book Analytic In: "Safety in ice hockey." Third volume, ASTMSTP 1341, By: Ashare AB, American Society for testing and materials.

Hayes D (1972). "The nature, Incidence, Location, and Causes of Injury in intercollegiate Ice Hockey." Book Analytic In: Application of Science and Medicine to Sport. By: Canadian Association of Sports Sciences, Springfield, Thomas.

Hodgson VR (1991). "Impact Standards For Protective Equipment." Book Analytic In: "Athletic Injuries to the Head, Neck, and Face." Second edition, By: Torg JS, pp 28-43, Mosby year book

Hodgson VR and Thomas LM (1972). "Effect of long Duration Impact on Head." SAE paper, NO 720956, proceedings of the sixteenth stapp car crash conference.

Honey CR (1998). "Brain Injury In Ice Hockey." Clinical Journal of Sport Medicine, 8:43-46, Lippincott-raven Publishers, Philadelphia.

Hurt HH and Thom DR (1985). "Laboratory tests and accident Performance of Bicycle Safety Helmet." 29th Annual Proceedings American Association for Automotive Medicine. Washington D. C.

Johnson GI (1989). "Development of an impact testing method for protective helmets." Book Analytic In: Safety in ice hockey. By: Castalid and Hoernr, USA.

Johnson RJ (1991). "Anatomy of the Brain and Its Coverings" Book Analytic In: "Athletic Injuries to the Head, Neck, and Face", second edition, By: Torg JS, pp 213-224, Mosby year book

Mellander H (1986). "HIC-the Head Injury Criterion: practical significance for automotive industry." Acta Neurochiurgica, Suppl. 36, pp 18-20.

Melvin JW, Lighthall JW, and Ueno K (1993). "Brain Injury Biomechanics." Book Analytic In: "Accidental injury: Biomechanics and Prevention." By: Nahum AM and Melvin JW. Springer-Verlag, pp 268-291, New York Inc.

Mills NJ and Gilchrist A (1997). " Pesponse of Helmets in Direct and Oblique Impacts." IJCrash, vol 2, No 1.

Morehouse CA (1989). "The ASTM F 1045-87 Standards Performance Specification for Ice Hockey Helmets." Book Analytic In: Safety in ice hockey. By: Castalid and Hoernr, pp 262-273, USA.

Nahum A, Ward C, Russell E, Adams S, and Schnieder D (1981). "Experimantal Studies of Side Impact to the Human Head." Proc. 25th Stapp Car Crash Conference, SAE 801301, 3806 (43).

Newman JA (1980). "Head Injury Criteria in Automotive Crash Testing." Society of Automotive Engineers, Inc., Biokinetics and Associates Ltd., Ottawa, Canada.

Newman JA (1983). "Tissue Stress Tolerances and Injury Indices." Biomechanical Assessment of Sports protective Equipment, 9th Congress of the International Society of Biomechanics, University of Waterloo, Waterloo, Ontario, pp 11-21.

Newman JA (1993). "Biomechanics of Human Trauma: Head Protection." Book Analytic In: "Accidental injury: Biomechanics and Prevention." By: Nahum AM and Melvin JW. Springer-Verlag, pp 292-310, New York Inc.

Northrip JW and McKinney L (1988). "Biomechanics analysis of sport", Book Analytic In: Science and Football. By: Reilly T, Lees A, Davids K, and Murphy WJ. E&F.N. Spon, USA

Nusholtz GS, Wylie EB, and Glascoe LG (1995). " Internal Cavitation in Simple Head Impact Model." Journal of Neurotrauma, 12(4): 707-14.

Ommaya AK and Gennarelli TA (1974). "Cerebral concussion and traumatic unconsciousness. Correlation of experimental and clinical observations of blunt head Injuries." Brain. 1974 Dec;97(4):633-54

Ommaya AK and Hirsch AE (1971). "Tolerances for cerebral concussion from head impact and whiplash in primates." J Biomech. Jan;4(1):13-21.

Ommaya AK, Rockoff D, and Baldwin M (1964). "Experimental Concussion." J Neurosurgery, 21: 249-265.

Ono K, Kikuchi A, Nakamura M, Kobayaslli H, and Nakamura N (1980). "Human Head Tolerance to Sagittal Impact: Reliable Estimation Deduced from Experimental Head Injury Using Subhuman Primates and Human Cadaver Skulls." 24th Stapp Car Crash Conference. Society of Automotive Engineers, 24:101-160.

Pearsall DJ, Wall RE, and Hoshizaki BT (2000). "Comparison of international safety standards for ice hockey helmets." Book Analytic In: "Safety in ice hockey." Third volume, ASTMSTP 1341, By: Ashare AB, American Society for testing and materials.

Reid SE and Reid SE Jr (1981). "Advances in sports medicine, prevention of head and neck injuries in football." Surgery annual, 13:251-70.

Rutherford GW and Miles RB (1981). "Overview of Sports Related Injuries to Persons 5-14 Years of Age." US Consumer Product Safety Commission, Washington, D.C.

Sabelli J and Morehouse CAD (2000). "Comparison of Ice Hockey Helmet Impact Attenuation Tests on Steel and Elastomeric Impact Surfaces." Book Analytic In: "Safety in ice hockey." Third volume, ASTMSTP 1341, By: Ashare AB, American Society for testing and materials.

Schneider K and Zernicke RE (1989). "Injury Tolerance of the Human Skull in Relation to Simulation of Impact Acceleration." Unfallchirurg, 92(2): 49-53.

Smodlaka VN (1984). "Medical aspects of heading the ball in soccer." The physical and sports medicine, Feb. vol.12, no.2, 127-131.

Spyrou E, Pearsall DJ, and Hoshizaki TB (2000). "Effect of Local Shell Geometry and Material Properties on Impact Attenuation of Ice Hockey Helmets." Black well Science, Sport Engineering.

Therrien RG (1984). "Eccentric Head-Bord Impact In Ice hockey." Sports Biomechanics. Proceeding of ISBS 1984, Del Mar, Calif., Research Center for Sports, p. 179-188.

Tysvaer A and Storli O (1981). "Association football injuries to the brain, a preliminary report" British J. Sports Med., vol.15, no.3, 163-166.

Vegso JJ and Torg JS (1991). "Field Evaluation and Management at Intracranial Injuries." Book Analytic In: "Athletic Injuries to the Head, Neck, and Face." Second edition, By: Torg JS, pp 225-231, Mosby year book

Yoganandan N, Pintar FA, Saces A Jr, Walsh PR, Ewing CL, Thomas DJ, and Snyder RG (1995). "Biomechanics of Scull Fracture." Journal of Neurotrauma, 12(4) 659-668.