A Comparative Analysis for the Measurement of Head Accelerations in Ice Hockey Helmets using Non-Accelerometer Based Systems

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Abstract

Head impact research involving accelerometer based sensors in sports helmets has been well documented over the past decade; however, the maximum number of players involved in the largest continuous study was 314 over 3 years. Practical methods for using accelerometer arrays present significant power management issues and a requirement for high resolution data during direct head impact injury. The objective of this study was to investigate a reliable and affordable method for measuring direct head impacts for large scale populations by using electromechanically activated force switches instead of accelerometers. An embedded micro-processor and software algorithm captured and calculated voltage activation of the force switches between 80-100 KHz. Laboratory studies conducted on a monorail drop tower using an ISO headform demonstrated the ability to correlate headform acceleration to algorithm reported acceleration. Impacts were performed on hockey helmets at 2.0, 2.5 and 3.0m/s for an aggregate percent difference of 8.9% at the front, front boss, side, rear boss and rear impact locations, respectively. The use of force switches in sensors affixed to sports helmets is viable at exceptionally low cost monitoring and analyzing reliable direct head impact linear acceleration.

Keywords: Electromechanical force switch, head impact, head injury, linear acceleration.

Introduction

Increased awareness of sports related minor traumatic brain injuries (mTBI) continue to gain interest from different groups in the medical, sports and parenting community. These concerns and discussions are the result of media reports of high profile player injuries in professional sport and the increased risk of head injury for many recreational athletes. At the recreational level, approximately 1.6-3.8million related mTBI incidents occur in the United States every year and in most cases minor incidences are not treated at the hospital [1]. The estimated medical and indirect costs of minor traumatic brain injury are reaching $60 billion annually [1]. While emergency facilities in North America collect data on admitted traumatic brain injuries (TBI) cases there is little statistical data collection on unreported mTBIs in athletes and non-research settings. Recent studies indicated a significant rate of under reporting of sports related mTBI due

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to many factors, including the simple inability of team staff to either recognize the signs and symptoms or witness the impact [2]. The majority of players involved in hockey and football are not college or professional athletes; however, there are over 3 million youth hockey players and approximately 5 million registered participants in football [3]. These recreational athletes have basic access to medical staff trained in concussion recognition and sideline injury assessment. A standardized user friendly measurement and assessment tool would facilitate the process between identifying potential head injuries, assessment, and return to play (RTP) criteria.

Since the 1940s, there has been an increasing amount of research into the forces that act on human tissue involved in various impulse and direct impact events. There are two types of forces that act on the head of athletes producing accelerations when either the head hits a stationary object or is struck by a moving object [4]. The forces applied to the head are measured and calculated as linear and rotational accelerations. Linear accelerations are measured and reported as “g”. Rotational accelerations (rad/s^2) are calculated from linear acceleration and were first introduced by Holbourn (1943) as a contributor to concussive type injuries [5]. Head injuries are the result of accelerations acting on the soft tissue which causes damage to the brain; regardless if the impact is applied directly or indirectly (impulse) to the head [6]. Mechanisms of injury as a result from linear and rotational accelerations are being proposed due to the inherent complex physical and physiological nature of the human brain from resulting mTBI. There is also a growing body of research; indicating the importance for understanding the long term consequences of repetitive impacts to the head and the possibility of more serious and detrimental injuries [7,8].

Recently, the use of instrumented sports helmets including the Head Impact Telemetry System (HITS™) (Simbex, Lebannon, NH), have allowed for detailed recording of impacts to the head in many research trials [10-12] leading to the recommendations to alter contact in practices and certain helmet design parameters. However, due to the high cost of the HITS system and complexity of the equipment, it is not a practical impact alert device for the general recreational population. Most recreational sports teams mandate the safe participation of athletes, rather than investment into instrumented helmets.

The objective of this study was to perform a comparative analysis of the dynamic impact response of a helmeted ISO headform on a monorail drop tower with the resulting output of a new kind of impact sensor for protective helmets. The application of a sensor to the wider market of untrained parents, team coaches and athletic therapist will be used for impact alerts and hit counter. It is proposed that such a device would act as a prompt to begin sideline assessment protocols for head injury while gathering data on the frequency and magnitude of impacts per player.

Methods

Limitations

Currently, instrumented helmets use accelerometers to measure peak linear acceleration and duration of accelerations to the head and often record between 15 and 50ms of data. Helmet accelerations generally exceed acceleration magnitudes experienced by the headform, accelerometers with a high dynamic range are required. Such accelerometers are usually expensive ($45-55 per axis) and require 3 accelerometers in each sensor package to measure linear acceleration in three-dimensions. The use of traditional accelerometers is considered impractical due to the high component cost, data management and high power consumption above 3mA/hr resulting in short battery cycles. These constraints reduce the practicality of a consumer device whereby it is likely that users can forget to recharge sensors prior to use. Finally, in consumer devices, simplicity and compatibility with familiar tools and techniques are important to ensure that devices are consistently used for their intended purpose.

Sensor Design Parameters

Due to the above mentioned technical and human factor constraints, a simple, practical, and affordable system was designed. A customized electronic component originally designed as a binary force switch replaces the accelerometers. Unlike accelerometers, binary force switches are exceptionally low cost and can be developed for a variety of multi directional uses and designed to activate at determined acceleration magnitudes and profiles.

Since the proliferation of smartphones in households for the purpose of communication, emails and texting; an internal study conducted as part of this research indicated over 75% of parents with children engaged in hockey or football used a smartphone as a primary means of communication. All smartphone devices have an embedded Bluetooth communication system to receive and transmit data at various ranges. Bluetooth systems are not dependent upon cell phone signals or coverage and can be interfaced by software applications on the smartphone. Therefore, a Class 1 Bluetooth device was chosen as the hardware communication method due to its simplicity, widely accepted standard and compatibility to interface with existing smartphones. Finally, all smartphones have considerable processing ability that exceeds many laptop computer devices of the past decade and can download and install custom software Applications known as "Apps".

Test Protocol

Using an impact sensor fitted with four (4) force switches, a Bluetooth transmitter, smartphone user interface and enabling electronics was used to correlate helmet accelerations with resultant headform accelerations. For impact testing, a monorail impact drop tower fitted with an ISO magnesium headforms and single uni-axial accelerometer secured at the headform's centre of gravity (CG). Impacts were directed onto a polyurethane covered impact board. A National Instruments NI 9174 data acquisition system was used to conduct all testing. Headform accelerations were sampled at 10 KHz.

A hockey helmet with expanded polypropylene (EPP) energy attenuating technology was impacted at 5 impact locations: front, front boss, side, rear boss, rear impact locations. For each impact location, three (3) velocities were chosen to represent an array of impact energy. The velocities were selected at 2.0m/s, 2.5m/s, 3.0m/s for 10 impacts at each condition for a total of 150 impacts. Individual impacts were conducted with time interval of no less than 120s in order to allow the helmet to sufficiently recover from each impact. The two dependent variables that were selected to be analyzed were measured peak resultant linear acceleration of the headform and calculated peak linear acceleration from the impact sensor.

Data Collection

The sensor was fitted with four (4) unidirectional orthogonally placed force switches in the X and Y. The X axis used 2 switches to measure front (+) and rear impacts (-), the Y axis used 2 switches to measure left (+) and right side impacts (-). Upon impact, the force switches activate with their respective axes and the on board electronics record the electronic voltage activations at each switch. It was discovered that the force switches have characteristic on and off voltage profiles when exposed to various accelerations during impacts to sports helmets. Different properties of the helmet shell, padding materials and axis of impact produced longer or shorter activation times and patterns.

Sensors were installed in ABS Nylon cases and attached to the crown exterior of the hockey helmet (Fig. 1.) using a 0.9mm PE foam adhesive transfer tape compatible to hockey helmet High density Polyethylene (HDPE) plastic. Several types and thicknesses of adhesives were tested to identify one with the least attenuation of impact energy. Sensors were affixed to the helmet crown due to the reduced likelihood of snag hazards or direct blows to the sensor. Sensors were tested according to CSA Z262.1-09 to identify any degradation to impact protection of the helmet.

Fig. 1. Position of impact sensors attached to exterior crown of standard hockey helmets.

Once the impact has occurred, the sensors micro-processor determines whether or not to transmit data based on a pre-determined threshold set of values that correspond to headform peak g for the particular helmet design being used. If the processor determines that the impact is too low to be of interest, it erases its memory and returns to a low power sleep function until activated. This threshold is set at approximately 50 g and varies according to impact location and helmet construction. This value was chosen due to the mechanical constraints of the force switch itself and as a design requirement to avoid spurious and erroneous sensor activation including small 10-30g impacts and impulses which are not the target activation points of interest for high risk impact events. The activation threshold was also chosen based on data from Gwin indicating 92% of hockey impacts were below 50 g [13] and 97% of football impacts were below 40 g linear acceleration recorded by Rowson [14].

Following an impact, the sensor processor determines if the impact is of interest and sends a request signal to the receiver smartphone based upon a set Bluetooth communication protocol. Each sensor is uniquely paired to the smartphone using the Bluetooth Media Access (MAC) address, a completely unique identifier from any other Bluetooth device for a maximum of 128 paired sensors to a single smartphone. Once the smartphone receives the request ID it transmits an acknowledge response back to the sensor whereupon it sends a data packet of the impact information. This data is then assessed using specific algorithms embedded in the smartphone Apps software particular to each type and model helmet. The algorithms provide an assessment of peak g linear acceleration and direction of impact in visual forms on the smartphone screen.

Results

The main effect of impacts directed through the helmet at five (5) impact locations on the monorail drop tower yielded no significant difference between the measured linear acceleration of the headform and calculated linear acceleration of the helmet ($F_{(14,128)}$=1.988, p=0.072). Significant difference was reported across the three (3) impact velocities ($F_{(14,128)}$=5.139, p<0.05).

Table 1 show the mean, standard deviation and percent difference comparing the measured linear acceleration (g) of the headform and the calculated linear acceleration (g) of the helmet across impact velocity and location.

Table 1—Peak linear acceleration (g) mean and standard deviation.[2]

	Front		Front Boss		Side		Rear Boss		Rear	
Velocity	Headform	Helmet	Headform	Helmet	Headform	Helmet	Headform	Helmet	Headform	Helmet
2 m/s	105.1 ±3.6	100.3 ±11.6	74.5 ±1.5	64.0 ±18.8	65.6 ±2.0	66.2 ±11.6	69.5 ±2.3	54.0 ±14.7	68.9 ±2.0	70.5 ±10.1
%diff.	4.6%		14.1%		0.9%		22.4%		2.5%	
2.5 m/s	151.1 ±7.8	163.2 ±111.6	90.3 ±3.6	89.4 ±14.9	93.6 ±3.5	97.3 ±7.5	85.8 ±2.4	67.6 ±13.4	96.4 ±1.8	106.1 ±6.5
%diff.	7.8%		1.0%		3.9%		21.2%		10.0%	
3.0 m/s	221.4 ±12.3	206.4 ±17.2	115.8 ±2.6	92.8 ±6.9	132.2 ±6.1	129.5 ±13.6	108.3 ±1.6	100.3 ±12.8	124.9 ±1.9	113.4 ±5.1
%diff.	6.8%		19.9%		2.0%		7.4%		9.2%	

The measured acceleration of the headform and calculated helmet acceleration demonstrated an aggregate percent difference across all locations and velocities of 8.9%. Aggregate percent difference across all impact locations at 2.0m/s was at 8.9%, 2.5m/s was 8.8% and at 3.0m/s was 9.0%. The sensor was able to predict the impact direction and approximate location on the helmet 100% of the time and was reported as front, side or rear locations through the software applications.

[2] In table 1, impact locations front, front boss, side, rear boss and rear impact locations. Percent difference is difference between meansured headform and calculated helmet linear accelerations (g).

Discussion

The purpose of this study was to demonstrate that the calculated linear acceleration of a hockey helmet measured by an externally mounted binary force switch is relatively similar to the measured linear acceleration of a headform dropped from a monorail drop tower. Due to the design limitations of the sensors, linear acceleration is the most reliable measurement variable to identify impacts directly to a hockey helmet. Figures 2 to 4 demonstrate the ability of a

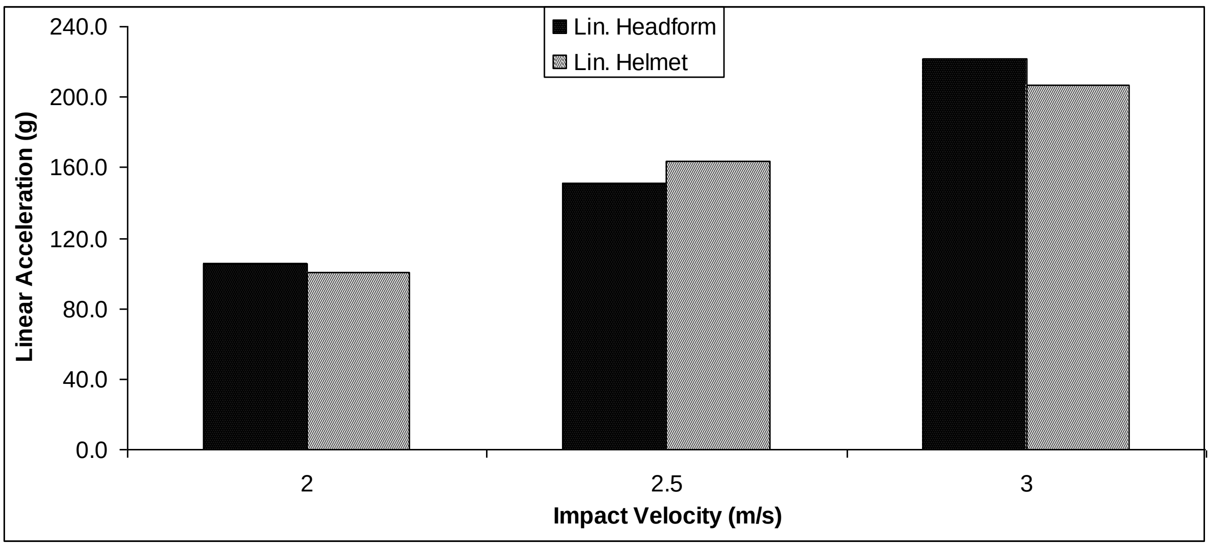

Fig. 2-- Front impact location across 3 impact velocities comparing the linear acceleration of a magnesium headform and the calculated linear acceleration of a helmet mounted impact sensor.

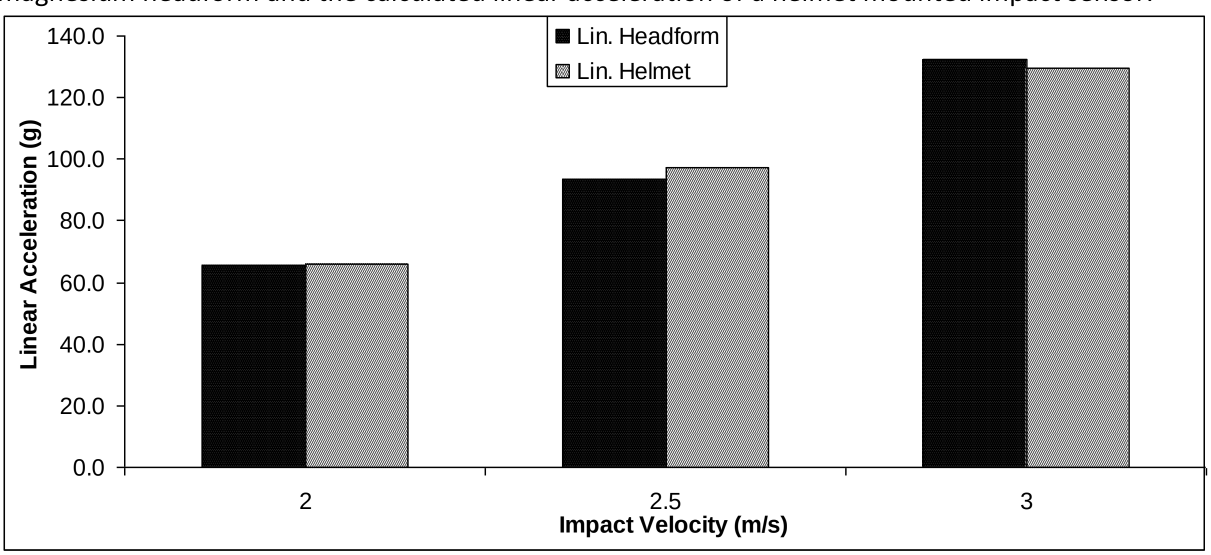

Fig. 3 -- Side impact location across 3 impact velocities comparing the linear acceleration of a magnesium headform and the calculated linear acceleration of a helmet mounted impact sensor.

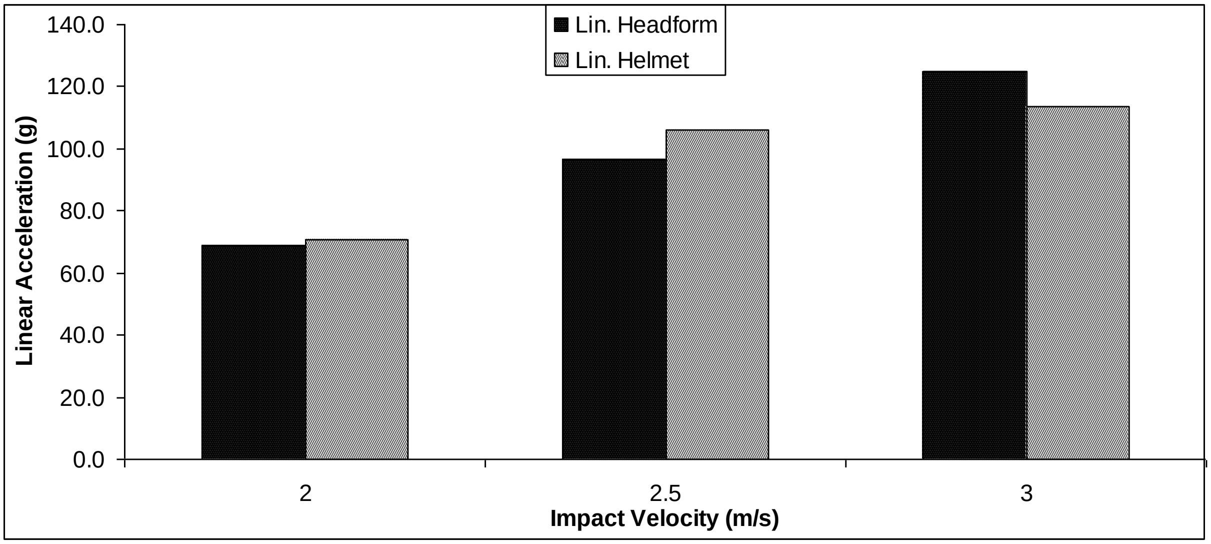

Fig. 4 -- Rear impact location across 3 impact velocities comparing the linear acceleration of a magnesium headform and the calculated linear acceleration of a helmet mounted impact sensor.

binary force switch in an ABS Nylon case mounted to the crown of a hockey helmet to be relatively accurate throughout a large energy range.

The sensors have been developed to trigger above 50g which represent 8% of impacts in hockey [13]. Further, the sensor trigger range is where the risk of head injury begins to increase to 25% at 66g and 50% at 80g [15]. The sensors perform better generally between the 50g and 90g range of impacts while mounted on hockey helmets, respectively. The application of these sensors to sports is best used as an impact alert device. With smartphone capability, there is access to different tools that may be used to provide information for understanding the risk of head injury to players immediately connected to the smartphone and supporting the sensor.

This study reports that low cost, non accelerometer based biomechanical sensors are a feasible concept with an acceptable rate of accuracy for mass market or large scale studies of direct helmet impact frequency and general magnitude alerting. From the data it shows that greater accuracy can be achieved at magnitudes below 90 g peak linear acceleration through software algorithm adjustments. Future research for this application would be the application of the sensor to in vivo subjects during recreational activities like hockey and football. Further, the use of this sensor to capture events in three dimensions as well as sensitivity to rotational acceleration components

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