THREE DIMENSIONAL ANALYSIS OF LINEAR AND ANGULAR ACCELERATIONS OF THE HEAD EXPERIENCED IN BOXING

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ABSTRACT

The mechanical response characteristics of a Hybrid III headform and mechanical neck to the left jab and left hook punches have been studied. Three Canadian amateur boxers participated in the study by delivering such blows to the head and neck of the Hybrid III which was solidly mounted to a concrete table. Angular and linear acceleration measures were taken simultaneously using a specifically designed nine accelerometer array mounted to the headform.

Both peak resultant linear accelerations and peak resultant angular accelerations were higher for the left hook punch than for the left jab punch. The results for both the left jab and the left hook demonstrated that neither punch was capable of causing head accelerations which were of concussive strength using published tolerance data.

The results demonstrate that the probability of receiving a concussive injury from a single punch of the types delivered in this study is quite low; however, the acceleration levels may be high enough to cause mild cerebral concussion in some instances.

INTRODUCTION

At the present time, only a few studies on boxing exist and very little is known about the forces developed from a boxing blow or the subsequent head acceleration effects following the blow. Duplication of the boxing ring environment presents both ethical and physical problems, thereby making research in this field very difficult.

As a suitable alternative, many researchers have used the surrogate headforms developed for automotive research instead of human subjects. Newman (1) employed amateur boxers and a Hybrid III head and neck system to assess the protective value of a variety of boxing helmets which were worn by the Hybrid III. The head and neck were mounted on a pedestal and a linear accelerometer was placed at the centre of gravity of the headform. The peak headform acceleration was 85 g with a punch impact velocity of 9.1 m/s. The assumption was made that all head accelerations remain purely translational and in the plane of measurement of the accelerometer. However, in general, when a boxer strikes the headform, the approach, the contact, and the followthrough tend to force the headform in several different directions. These directional accelerations would not be detected or measured by the uniaxial accelerometer and therefore, the resultant headform accelerations may actually be higher.

A study by Kozey (2) accounted for this by using a triaxial accelerometer mounted at the centre of gravity of a Hodgson-WSU headform which was in turn mounted securely to an automotive spring in order to simulate the neck response of the boxer being struck. Both amateur and professional boxers participated in the study. Their weights ranged between 132 to 181 pounds (lightweight to heavyweight) and the mean peak triaxial acceleration was 84.2 g with a mean GSI of 162.2. The author reported a variability between boxers and between individual punches as high as 30 percent in some cases. As well, the effect of a freely moving "neck" upon the resultant linear accelerations is not known. It may be suggested that this is not a true representation of a boxer's neck since most boxers undergo intensive training in order to develop a very stiff neck, probably in an effort to reduce the amount of head movement which may occur following a blow.

The most recent study involving measurement of boxing blows with anthropomorphic test dummies was done by Schwartz et al. (3) using a Hybrid II headform mounted on a universal joint which pennitted movement about three orthogonal axes. The Hybrid II headform was also covered with a 5 cm thick layer of medium density foam as well as a 1 cm layer of Plastizote to simultate the friction properties of the skin. The overall damping effect that this would have on the acceleration signal is not known. Following application of the padding, the apparatus was then mounted onto a stiff steel column which was bolted directly to the floor. Fourteen black belt karate subjects were then asked to either punch or kick the headform apparatus. The headform acceleration results from the boxing punches are not well reported, however, the authors mention that peak linear accelerations of 90 g were recorded on several occasions and one value of 120 g was also recorded.

It is apparent from the boxing data presented to date, that there is not enough evidence to prove that linear acceleration effects alone are sufficient to cause acute head injury in boxing. However, the linear acceleration levels may be enough to cause some form of head injury if they were coupled with significant angular acceleration effects (4).

Therefore, adequate surrogate test devices must be capable of measuring both linear and angular acceleration simultaneously. Unfortunately, only a limited amount of research has been done in the area of angular acceleration measurements. This is largely due to limitations in the development of accurate angular acceleration measurement systems (5).

This paper describes the development of a system for measuring three dimensional linear and angular accelerations of the head experienced in boxing and the results of a punching study using three Canadian amateur boxers. The tolerance limits selected from the literature were 200 g for linear head acceleration (6) and 4500 rad/s/s for angular head acceleration effects (7). The results of the punch study will be compared to these tolerance values and discussed in light of difficulties encountered when attempting to simulate the boxing ring environment.

METHODOLOGY

All impacts were directed towards a Hybrid III headform with an accompanying Hybrid III mechanical neck. A boxing helmet was placed on the headform for all punching trials. In addition to the head and neck model, an accelerometer mounting system was located at the back of the skull. This device was fabricated from aluminium, steel, and polyvinylchloride (PVC). The aluminium skull plate was tooled to resemble the original Hybrid III rear skull plate. Two protrusions were made in the skull plate so that it was possible to attach two steel rods which travelled from the posterior of the dummy head to the front of the dummy head in direct line with the centre of gravity of the headform. The location of the centre of gravity was taken directly from the reference markings found on the magnesium Hybrid III skull. A PVC cube was then threaded onto the end of each steel rod such that the cube was located over the centre of gravity of the headform. A third cube was mounted directly onto the back of the skull plate and oriented to be directly in line with the centre of gravity as well. The cubes located at the end of the steel rods were fashioned to rest upon the latex skin of the dummy headform. This was done to reduce any high frequency oscillations which might have occured following impact.

A triaxial accelerometer was located at the head centre of gravity and upon each arm was mounted a pair of uniaxial accelerometers oriented such that they were positioned orthogonally along the axes originating from the head centre of gravity. Another pair of uniaxial accelerometers was mounted on the rear cube in a similar manner (see Figure 1). This produced the typical 3-2-2-2 configuration with the resultant linear acceleration calculated from the triaxial accelerometer and the angular accelerations about all three planes of motion calculated using the equations:

$$\alpha_x = \frac{(a_{z1} - a_{z0})}{2l_1} - \frac{(a_{y3} - a_{y0})}{2l_3}$$

$$\alpha_y = \frac{(a_{x3} - a_{x0})}{2l_3} - \frac{(a_{z2} - a_{z0})}{2l_2}$$

$$\alpha_z = \frac{(a_{x1} - a_{x0})}{2l_1} - \frac{(a_{y2} - a_{y0})}{2l_2}$$

thus, the angular acceleration about each axis is calculated by taking differences between pairs of accelerometers within the nine accelerometer cluster and dividing by the distance between the two accelerometers (1). Figure 2 illustrates the axis system for measuring translational and rotational accelerations in relation to the Hybrid III headform.

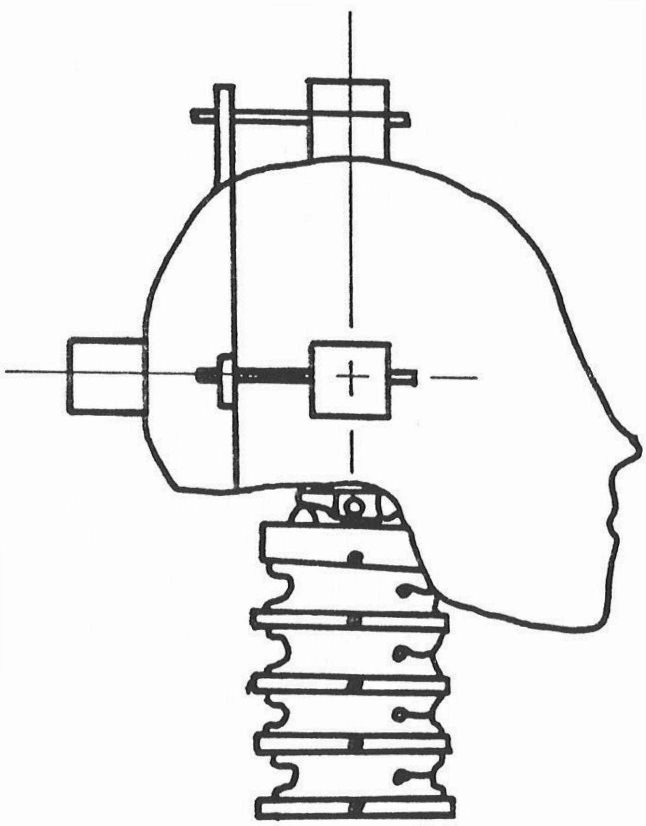


Figure 1: Hybrid III Headform with Rear Skull Plate

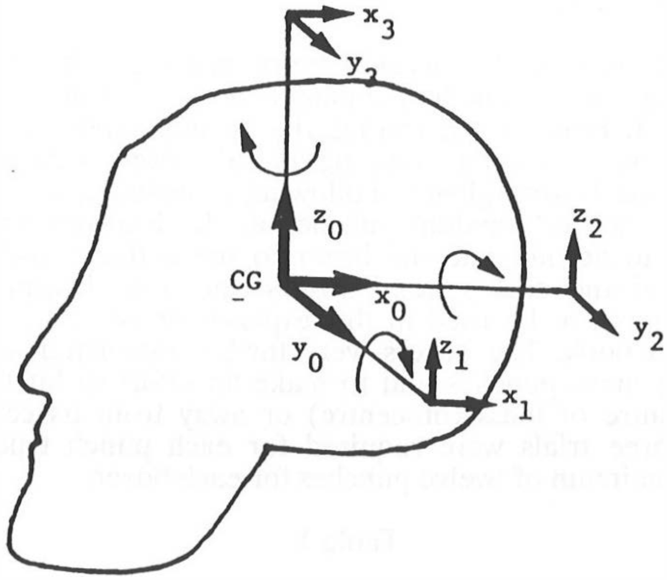


Figure 2: Accelerometry System Mounted on the Hybrid III Headform

The advantages of the use of this nine accelerometer array for impact biomechanics have been explained previously by Padgoankar et al. (8). The above equations were written into a computer program which performed all linear and angular acceleration calculations and provided graphical output for each trial.

VALIDATION OF THE HEAD IMPACT MODEL

Prior to testing, the head accelerometry system was properly validated to ensure accuracy during testing with the boxers. The validation procedure included comparing the physical parameters of the new headform apparatus with those of the original Hybrid III headform as well as validation of the nine accelerometer system.

The results of this validation indicated that the new rear skull plate caused minimal movement of the original head centre of gravity (3 mm) and the addition of the rear skull plate and boxing helmet caused a net 29 % increase in the mass moment of inertia. The implications of this increase moment of inertia shall be discussed relative to the angular acceleleration findings.

Following the validation tests, the angular acceleration measurement system was found to be very accurate, indicating its suitability for measuring headform accelerations generated from boxing punches. A detailed discussion of these validation tests and validation results may be found elsewhere (9).

BOXING PUNCH STUDY

Following validation of the accelerometry system, a boxing study was conducted using three Canadian amateur boxers. Subject data may be found in Table 1. Prior to any testing, the boxer's hands were taped with the regulation length of hand wrapping and all boxers used the same pair of Everlast 12 oz boxing gloves. Following a warmup, the boxers were allowed to take several random punches at the instrumented headform. This was done to accomodate the boxer to the stiffness and response of the dummy head and neck system. Boxers were then instructed that two punches were going to be used in this experiment and they were the left jab and the left hook. The boxers were further instructed to contact the headform using these punches and to make an effort to hit the headform either at its centre of mass (on-centre) or away from its centre of mass (off-centre). Three trials were required for each punch type and punch location for a minimum of twelve punches for each boxer.

Table 1

Subject Data

Subject	Age	Mass (kg)	Boxing Division
WT	17	57	Featherweight
GJ	20	60	Lightweight
LL	21	100	Superheavyweight

All subjects were instructed to stand in a boxing ready position and upon command from the experimenter, they were instructed to hit the headform with the required punch. Data collection was initiated by the triggering of a reflected light beam. Once the beam was broken by the boxer's fist, all nine accelerometer signals were A/D converted at a rate of 2300 hz per channel using the Watscope Data Acquisition System. Once the trial window was selected, the HYB3D program removed bias, calibrated the signal, filtered each channel at 150 hz using a 4th order Butterworth low pass digital filter and calculated the resultant linear and angular headform accelerations.

RESULTS

LINEAR HEADFORM ACCELERATIONS

Typical resultant linear headform accelerations for the left jab and left hook are presented in Figures 3 and 4. These plots are considered to be typical of all the data observed. A minimum of three trials per condition were averaged for each subject and the mean linear headform accelerations for all subjects are presented in Table 2. Across all subjects, the left hook accelerations were greater than the left jab acceleration values. Average headform acceleration across subjects was

22 g for the left jab and 21 g for the off centre left jab. The average on centre left hook headform acceleration was 44 g while the off centre average headform acceleration for the left hook was 58 g.

Comparison of the on centre and off centre punch locations showed only a marginal increase in on centre headform accelerations for the left jab while the left hook showed higher mean headform accelerations for the off centre punches.

Table 2
Mean Peak Linear Resultant Headform Accelerations (+/- SD) For the Left Jab and Left Hook Punches

Subject	Left Jab		Left Hook	
	On Centre	Off Centre	On Centre	Off Centre
	g	g	g	g
GJ	22.1 (2.0)	24.4 (1.6)	35.8 (7.6)	58.6 (3.1)
LL	25.4 (3.9)	23.9 (.9)	62.2 (5.9)	78.4 (9.8)
WT	16.9 (2.7)	13.4 (2.5)	32.8 (6.0)	35.5 (6.5)
Across Ss	21.5 (4.6)	20.6 (5.6)	43.6 (15.6)	57.5 (18.9)

Angular acceleration measurements were taken in all three planes of motion, frontal, sagittal, and transverse. Figures 5 and 6 show typical resultant angular headform accelerations as measured by the accelerometer array. The planar angular acceleration patterns remained quite consistent both within and between all subjects. Mean resultant angular acceleration data (+/- 1 SD) are shown in Table 3.

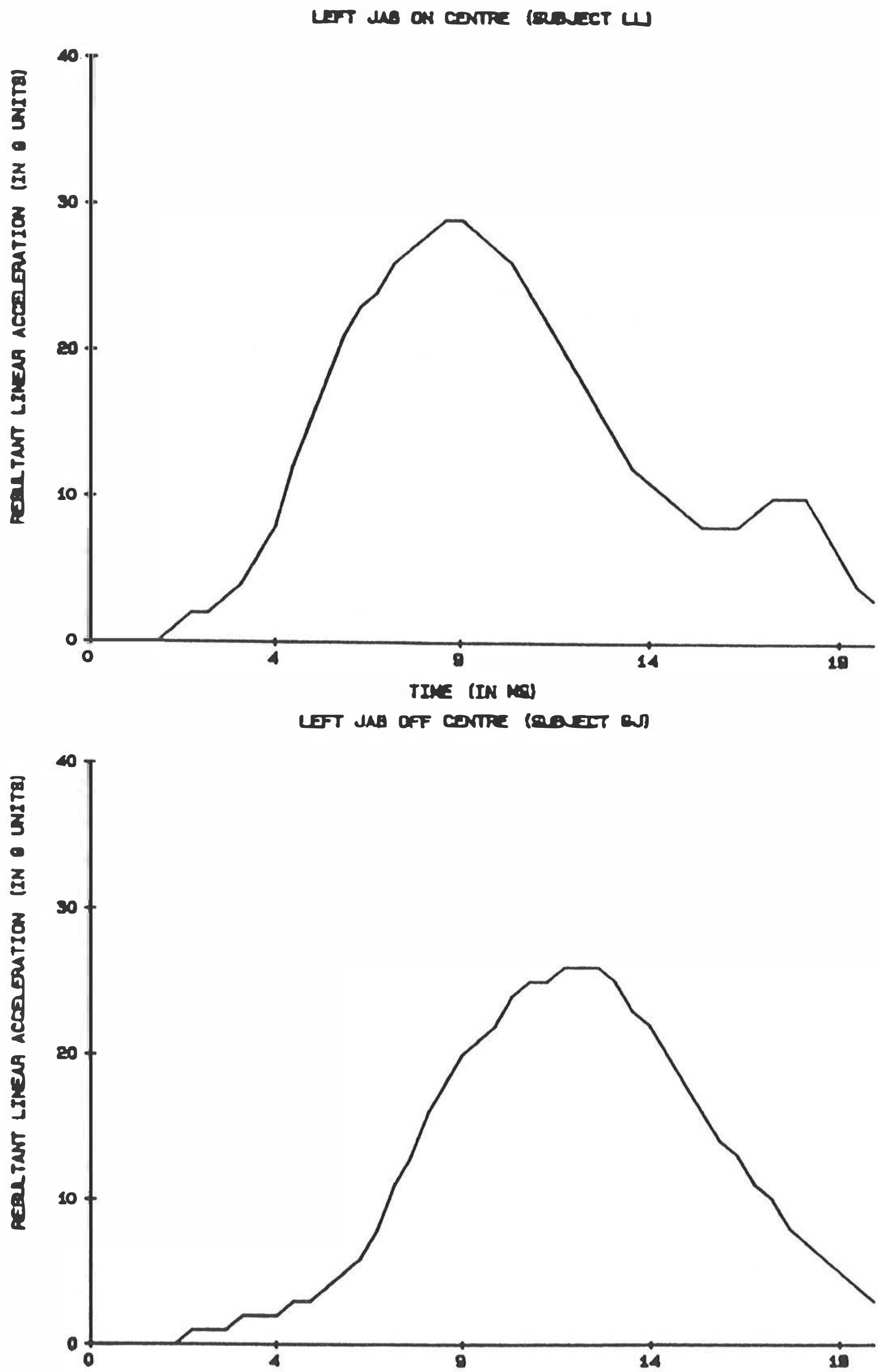
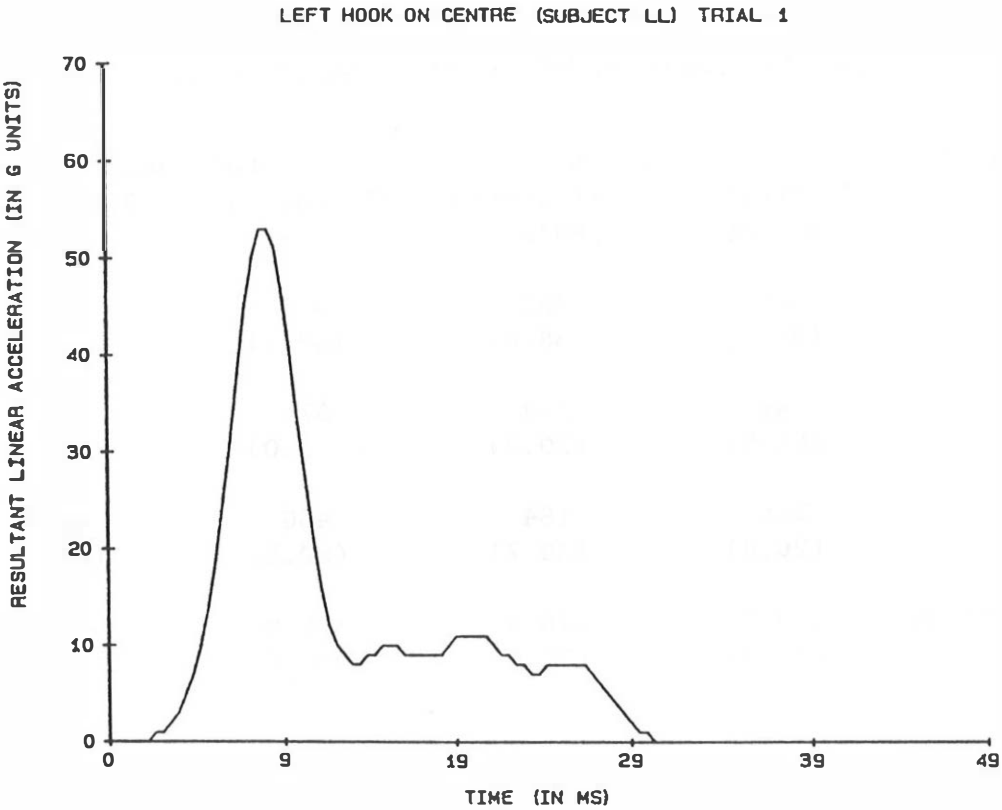


Figure 3: Linear Headform Accelerations for the Left Jab.



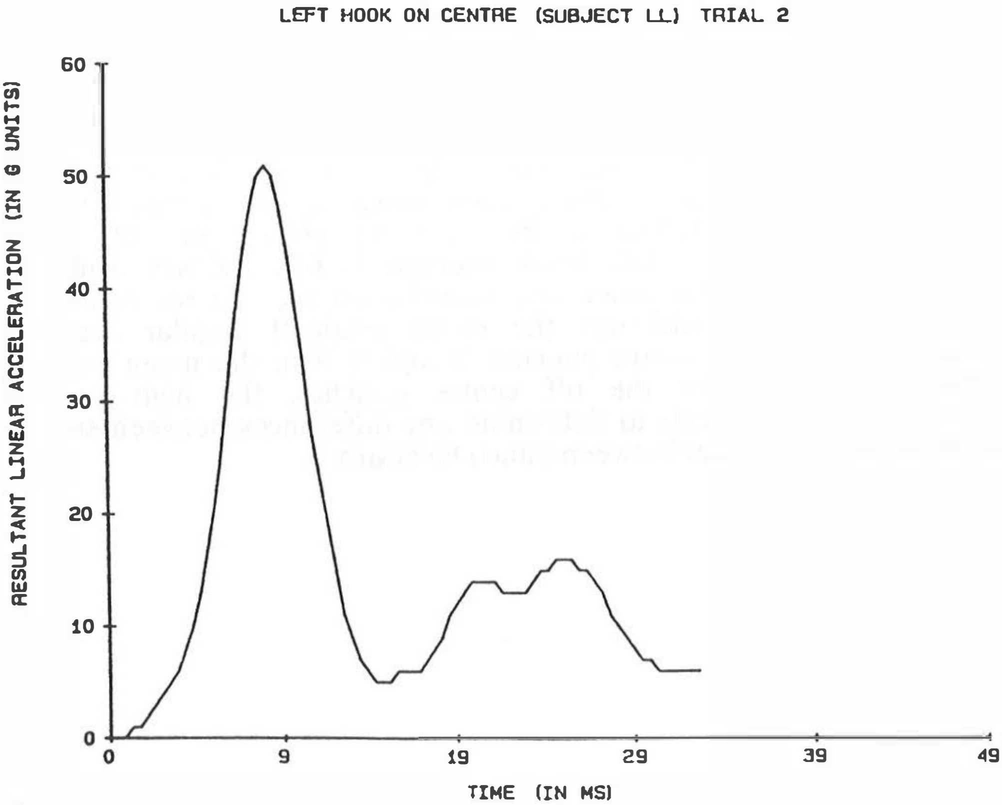


Figure 4: Linear Headform Accelerations for the Left Hook.

Table 3
Mean Peak Angular Resultant Headform Accelerations (+/- SD) For the Left Jab and Left Hook Punches

Subject	Left Jab		Left Hook	
	On Centre rad/s/s	Off Centre rad/s/s	On Centre rad/s/s	Off Centre rad/s/s
GJ	311 (35.1)	352 (38.8)	497 (109.1)	697 (33.9)
LL	344 (55.8)	324 (20.3)	874 (119.0)	859 (36.8)
WT	206 (26.0)	154 (39.7)	450 (80.5)	497 (65.7)
Across Ss	292.7 (72.2)	276.6 (97.5)	675.9 (230.6)	644.1 (159.8)

As with the resultant linear data, the resultant angular accelerations for the left hook are higher than those for the left jab. Grouping across subjects shows a large difference between the two punches. Average resultant angular headform acceleration for the left jab punches are 293 rad/s/s and 277 rad/s/s for on centre and off centre punches respectively. The left hook punches have much higher average resultant angular headform accelerations for both on centre and off centre punches. The on centre left hook average is 676 rad/s/s while the average resultant angular headform acceleration for the off centre left hook is 644 rad/s/s. Although the mean resultant angular headform acceleration for the on centre punches is higher than the mean resultant angular acceleration for the off centre punches, the high standard deviations make it difficult to determine any differences between subjects and any actual differences between punch locations.

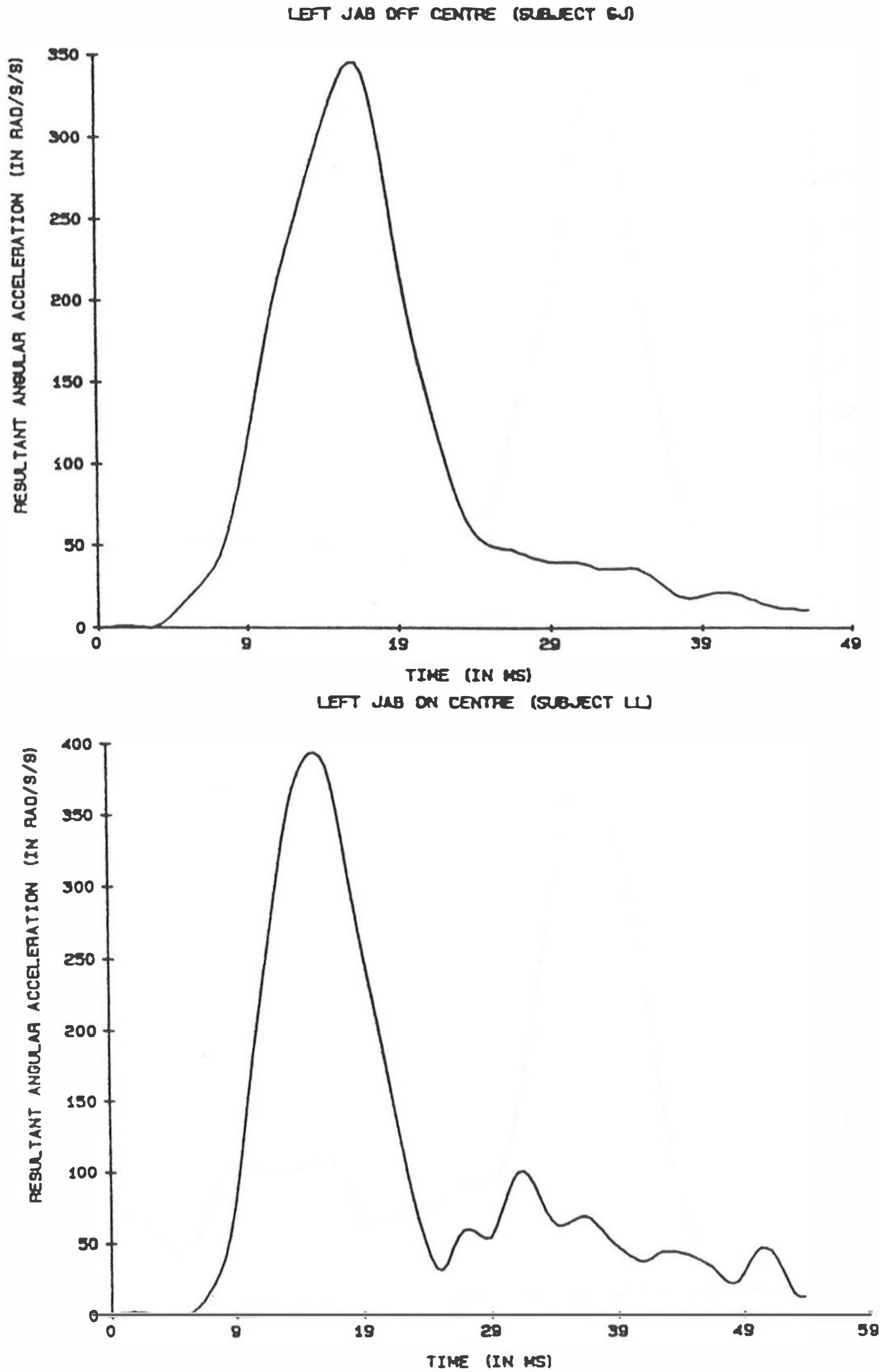
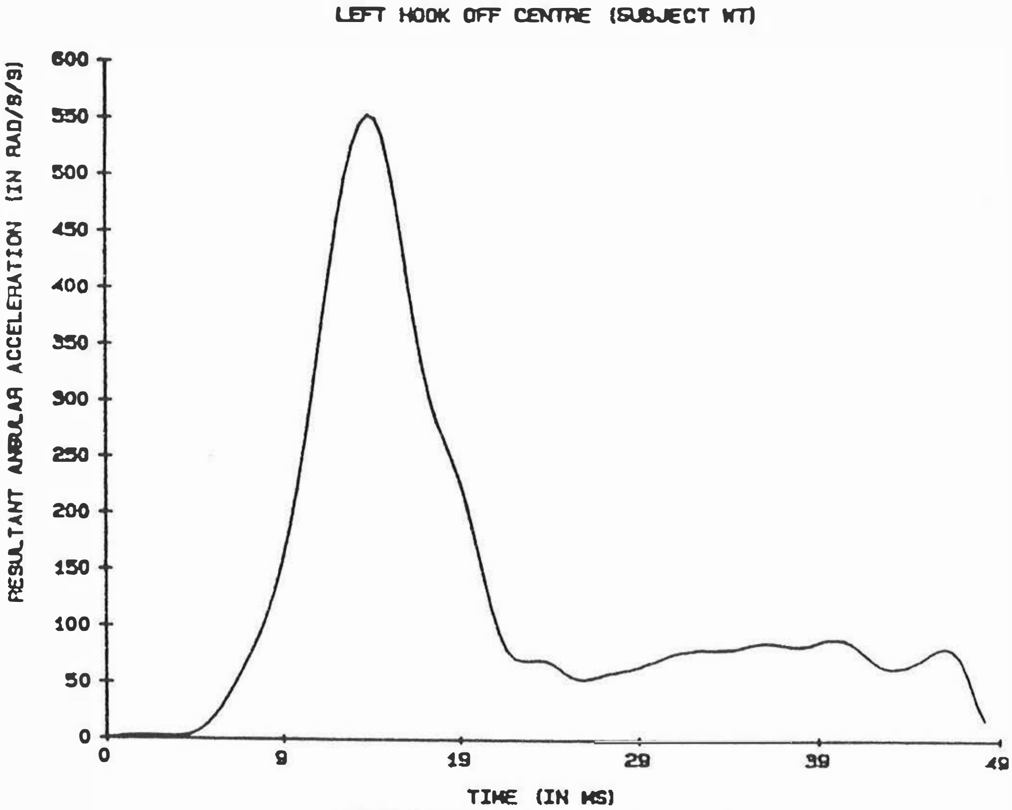


Figure 5: Resultant Angular Accelerations For the Left Jab.



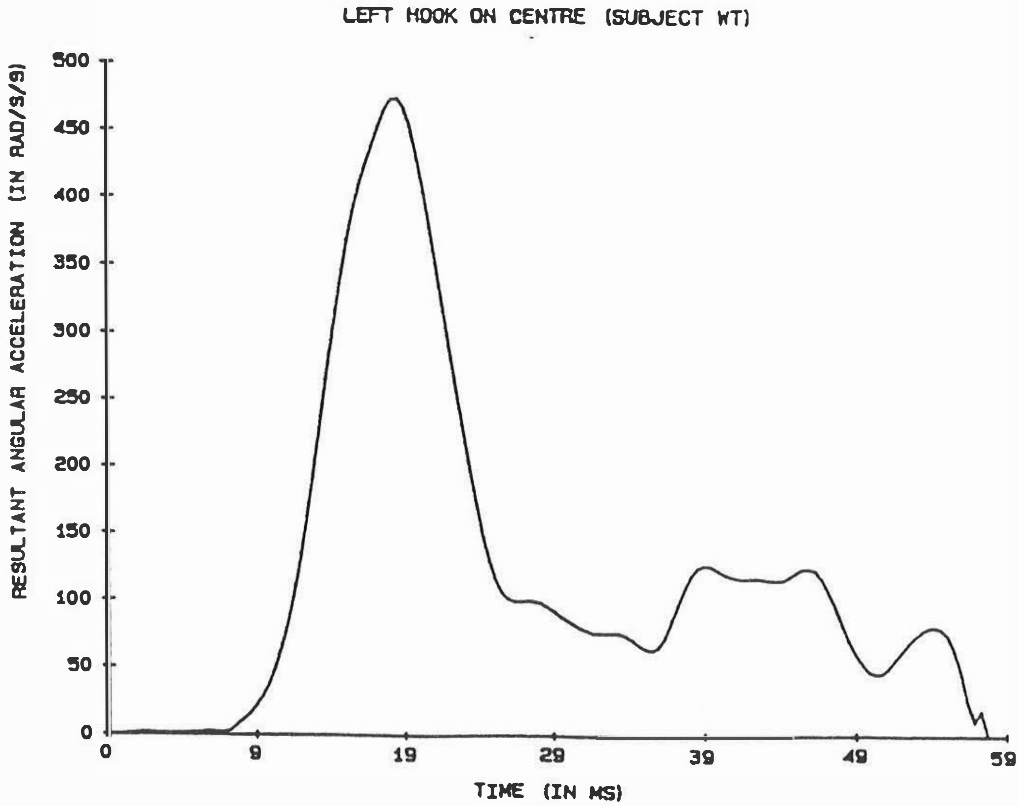


Figure 6: Resultant Angular Accelerations For the Left Hook.

DISCUSSION

The complex functioning of the brain and the diverse nature of head injury makes it very difficult to understand the different mechanisms and effects of head inju y. Perhaps the greatest problem is in the design of kinematic experiments which will not interfere with the physical system being monitored, yet provide complete and accurate results that will desc ibe the major effects of head impact. This becomes nearly impossible when considering the fact that there may be several different inju y mechanisms present during any given impact.

The development of human surrogates for impact testing has greatly improved this situation by providing accurate and repeatable kinematic results for a variety of impact situations and a variety of experimental techniques. Care should be taken however, in interpreting the results of these tests because much of the misunderstanding with regards to impact kinematics stems from the assumptions which were generated because of experimental techniques that do not (or cannot) accurately measure the parameters required.

This is not to say that the human model simulations should not be used for head injury research. Rather, the limitations of using these impact models and the limitations of the experimental techniques used should be well explained with specific reference to the human system which is being modelled. This step in the deductive reasoning process will then allow for a much better explanation of the potential mechanisms which may or may not be present during head impact.

The present study was undertaken to examine the linear and angular accelerations experienced by the head when subjected to selected punches that are typically used in a boxing match. The punches were delivered to a Hybrid III accelerometer array from which both resultant linear and resultant angular acceleration measures were calculated.

Prior to the actual testing, the battery of tests performed on the fabricated three dimensional accelerometry system provided consistent evidence with regards to the system's validity and accuracy in recording head impacts.

It was initially suggested in this study that the head accelerations seen by the Hybrid III headform during typical boxing punches would exceed both the linear (6) and angular (7) acceleration limits previously established for human tolerance to concussion or brain injury. Both the resultant angular headform acceleration values and the resultant linear headform accelerations are too low to support this hypothesis.

As has been suggested, the interpretation of this result is dependant upon a thorough knowledge of the limitations of the test equipment and the difficulties in the development of accurate measurement systems.

Table 4 demonstrates some of the factors which may limit the interpretation of the headform acceleration levels produced during this study.

Table 4
Factors Affecting Headform Acceleration

	Net Affect
Helmeted Headform	Decrease Headform Acceleration
Increased Mass Moment of Inertia	Decrease Headform Acceleration
Neck Stiffness	Decrease Headform Acceleration
Non-maximal Punch Effort	Decrease Headform Acceleration

Although the helmet functions largely to prevent facial lacerations, it does possess some impact absorbing capabilities. The net increase in the mass moment of inertia due to the boxing helmet and the rear skull plate will result in a net decrease in angular acceleration because the net moment about the point of rotation in the neck is equal to the product of the moment of inertia and the angular acceleration. Therefore, it the inertia tends to increase, then the angular acceleration will tend to decrease given the net joint moment. This may reduce overall angular acceleration that the headform experiences during a boxing punch.

It is also evident from this table that one of the major limitations in head injury research is the development of an accurate model for head impact which duplicates the human response as closely as possible. The situation created in the laboratory was one of ideal conditions for the boxer. Actual bouts involve mainly glancing blows, deflected blows and off balance blows. It is only if the boxer is seriously fatigued or slightly stunned that he will be open to receive a punch which would be identical to those seen in this study.

A better approach may be to directly record the boxing events as they occur in the ring as was done by Chamouard et al. (10). Their study involved a pair of boxers who were fitted directly with headgear designed to measure both linear and angular accelerations and asked to box. Although this approach does present many physical limitations, ie. restriction to physical movement, it may be a better representation of the actual boxing environment.

In conclusion, it can be seen that although boxing does possess inherent danger in the form of high levels of head acceleration, the probability of receiving an acute dangerous blow does not appear to be that great. The results of this study tend to suggest that cumulative trauma, in the form of repeated sub-concussive level punches, may be a factor in the development of neurological impairment in boxers. Therefore, it may be the gradual development of a series of minute anatomical disruptions, through constant low level punching, that predisposes a boxer to knockouts or even sudden death in the boxing ring.

Unfortunately, techniques for the accurate measurement of the head accelerations experienced in boxing are o ly currently being developed and little is known with regards to human tolerance to the cumulative trauma experienced during boxing impacts. It is hoped that the apparatus developed for this study may be used for further boxing research involving a greater number and wider variety of boxers.

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