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Concussion in Professional Football: Comparison with Boxing Head Impacts—Part 10

OBJECTIVE: This study addresses impact biomechanics from boxing punches causing translational and rotational head acceleration. Olympic boxers threw four different punches at an instrumented Hybrid III dummy and responses were compared with laboratory-reconstructed NFL concussions.

METHODS: Eleven Olympic boxers weighing 51 to 130 kg (112–285 lb) delivered 78 blows to the head of the Hybrid III dummy, including hooks, uppercuts and straight punches to the forehead and jaw. Instrumentation included translational and rotational head acceleration and neck loads in the dummy. Biaxial acceleration was measured in the boxer's hand to determine punch force. High-speed video recorded each blow. Hybrid III head responses and finite element (FE) brain modeling were compared to similarly determined responses from reconstructed NFL concussions.

RESULTS: The hook produced the highest change in hand velocity (11.0 ± 3.4 m/s) and greatest punch force (4405 ± 2318 N) with average neck load of 855 ± 537 N. It caused head translational and rotational accelerations of 71.2 ± 32.2 g and 9306 ± 4485 r/s^2. These levels are consistent with those causing concussion in NFL impacts. However, the head injury criterion (HIC) for boxing punches was lower than for NFL concussions because of shorter duration acceleration. Boxers deliver punches with proportionately more rotational than translational acceleration than in football concussion. Boxing punches have a 65 mm effective radius from the head cg, which is almost double the 34 mm in football. A smaller radius in football prevents the helmets from sliding off each other in a tackle.

CONCLUSION: Olympic boxers deliver punches with high impact velocity but lower HIC and translational acceleration than in football impacts because of a lower effective punch mass. They cause proportionately more rotational acceleration than in football. Modeling shows that the greatest strain is in the midbrain late in the exposure, after the primary impact acceleration in boxing and football.

KEY WORDS: Boxing, Concussion, Impact biomechanics, Sport equipment testing, Sport injury

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Closed head injury is an occupational hazard of many sports, and specifically in boxing and football. Participants in both sports are at risk for sustaining concussions ("dinged," "knocked out," cerebral concussion, MTBI). Zazryn et al. (101) found 107 injuries to 427 professional boxing participants, 89.8% of the injuries were to the head, neck and face with 15.9% concussions. For amateur boxers, the incidence of concussion is 4.0% to 6.5% (19, 38, 98). The difference in concussion rates between professional and amateur boxing may be due to differences in safety gear, and there have been recommendations to analyze professional and amateur boxers' injury rates separately (41).

The clinical picture of more severe brain injury is different in football and boxing (60–64, 75, 88, 89). The pattern of brain injuries in boxing has been extensively studied (2, 6, 7, 12, 13, 15, 18, 20, 22, 25, 26, 29–31, 36, 37, 44, 47, 51, 67–69, 72–74, 82, 85, 98). The pattern of brain injury in professional football has been recently studied and reviewed (60–64).

The medical literature is clear on the difference in the acute phase. Boxers are much more likely to develop subdural hematomas and brain-injury deaths than professional football players (40, 42, 66, 88, 89). Boxers are also prone to develop, over the long term, a characteristic pattern of chronic brain injury (chronic encephalopathy of boxers, pugilistic dementia, "punch-drunk") that has never been reported in professional American football players (12, 39, 47, 75, 87, 89).

The biomechanical forces affecting the brain of professional football players have been recently reported as part of the ongoing studies of the NFL (58, 59, 64, 93–95). This study of the biomechanical forces in boxing was undertaken to help explain the similarities and differences between the clinical overview of brain injuries in the two sports. A longer term goal of this research is to study the effectiveness of protective headgear and sport equipment, including helmets in football and headgear and gloves in boxing. An understanding of the biomechanical forces causing injury is the first step in addressing improved protection. Boxing gloves and headgear are currently required in amateur boxing to prevent head injury (98). The equipment may reduce some injuries, but it does not eliminate the risk of knockouts (101).

The effectiveness of boxing safety equipment has been addressed by Schmidt-Olsen et al. (81) in a three-year study of amateur boxers in Denmark. No decrease in injuries was found with an increase from 8- to 10-ounce gloves, unlimited hand-wrap and use of helmets for heavier boxers. The lack of other data on this topic leaves boxing officials, athletes and trainers uncertain as to what specific safety equipment is most effective and what areas of improved safety needed additional study.

An improved understanding of the mechanisms of brain injury and biomechanics of head responses in amateur boxing is needed to lay the foundation for improvements in the effectiveness of protective equipment in boxing. This study of boxing and the biomechanics of the punch involves several factors, including how much force is exerted during a punch type, how that force is transferred to the opponent's head, how the opponent's head responds to the punch and how the opponent's head responds to different punches.

Punch Forces in Boxing

The biomechanics of punches has been studied using surrogates simulating the opponent (3, 83, 99). Atha et al. (3) and Smith et al. (84) analyzed different surrogates that have been used, including a ballistic pendulum, a uniaxial strain gauge platform, instrumented punching bags, water-filled elastic bags and boxing dynamometer. A consideration when choosing a surrogate is its ability to mimic the human body in both shape and impact response.

Atha et al. (3) used a single boxer and an instrumented ballistic pendulum to evaluate a single straight punch. The professional boxer punched the surrogate with 4096 N, which the author estimated translated into 6320 N of force to a human head. This force produced peak acceleration for 53 g's on the 7 kg ballistic pendulum.

Joch et al. (34) placed 70 boxers into one of three categories, including 24 elite, 23 national and 23 intermediate boxers. The force of straight right punches was measured with a water-filled punching bag fit with a pressure transducer. The average maximum punch force was 3453 N, 3023 N and 2932 N, respectively. In addition to the lack of biofidelity, there was a need to stabilize the surrogate and de-gas the bag, which made testing cumbersome (84).

Smith et al. (84) developed a boxing dynamometer to measure punch force. Twenty-three boxers were sorted into elite, intermediate, and novice boxer categories. Boxers were instructed to punch the head region of a pear-shaped bag mounted to a wall. Boxers threw straight punches using both their lead and rear hands. Punches were thrown singularly and in two and three punch combinations. The elite boxers had a mean rear-hand punch force of 4800 N and a front-hand punch force of 2847 N. The intermediate boxers' rear and front hand punch forces were 3722 N and 2283 N, respectively, and the novice boxers' mean rear and front hand punch forces were 2381 N and 1604 N (84). The researchers developed a surrogate, but the faceless pear-shaped device lacked the human response provided by the neck.

Walilko et al. (97) recently studied the biomechanics of straight punches to the jaw causing translational and rotational head acceleration. This was a precursor study to the present investigation. Seven Olympic boxers from five weight classes delivered 18 straight punches to the compliant face of the Hybrid III dummy. Instrumentation included hand acceleration and pressure distribution on the jaw. The punch force averaged 3427 ± 811 N, hand velocity 9.14 ± 2.06 m/s and the effective punch mass 2.9 ± 2.0 kg. The jaw load was 876 ± 288 N. The peak translational acceleration was 58 ± 13 g, rotational acceleration 6343 ± 1789 r/s^2 and neck shear 994 ± 318 N. They found that boxers deliver straight punches with high impact velocity and energy transfer to head rotation. The severity of the punch increased with weight class primarily due to a greater effective mass of the punch.

Assessing Head Injury Risks

Following the methods of Pellman et al. (58), the risk of head injury was determined by the methods of Gadd (21) and Hodgson et al. (32) using a human surrogate that has biofidelity in its impact response. Biofidelity reflects the ability of the surrogate to simulate the essential biomechanical characteristics of the human impact response. The Hybrid III dummy used in this effort is currently the most advanced, validated biomechanical surrogate, particularly for the head and neck areas. Sensors placed in the surrogate collect biomechanical data that are related to risk of injury. Previous studies have developed criteria to estimate the risk of head injury from various impacts (24, 28, 55).

One of the earliest head injury criteria was based on research conducted at Wayne State University (46). By investigating the relationship between the level of acceleration and duration of the impact, the Wayne State Tolerance Curve (WSTC) was developed (27). Impacts to the head that were lower in acceleration required a longer pulse duration to cause the same injury as those higher in acceleration.

From this initial research, Gadd (21) expanded the analysis by including other human tolerance data from Eiband (17) and plotted the effective or average acceleration versus duration of impact on a log-log scale. The result was a straight line that had a slope of –2.5. Based on this result, the Severity Index (SI) was developed relating head acceleration to risk of injury:

$$\mathrm{SI} = \int^{T} a(t)^{2.5}\mathrm{dt}$$

where a(t) is the resultant translational acceleration at the head center of gravity (cg) and T is the duration of the acceleration. SI depends on the time history of the resultant translational acceleration. From the existing data, an SI tolerance of 1000 was established.

Versace (90) presented a new method for determining a head impact injury that took the SI one step further by optimizing the formula over the duration of impact. The final result was the Head Injury Criterion (HIC):

$$\mathrm{HIC} = \{(t_2 - t_1)[\int_{t_1}^{t_2} a(t)\mathrm{dt}/(t_2 - t_1)]^{2.5}\}_{\max}$$

where t_1 and t_2 are determined to give the maximum value to the HIC function and a(t) is the resultant translational acceleration of the head cg. In practice, a maximum limit of $T = t_2 - t_1 = 15$ milliseconds is used.

Risk of head injury is calculated from accelerations at the head cg in three orthogonal directions. The resultant acceleration is calculated from these measurements and is used to determine HIC. As summarized by Walilko et al. (97), the final value provided a maximum acceptable value. The US delegation to Working Group (WG) 6 provided an estimate of the percent of the adult population expected to experience a life-threatening brain injury (AIS 4) for various HIC levels due to frontal head impacts (70). The delegation's best estimate is that 16% of the adult population would experience a life-threatening brain injury at a HIC level of 1000. In a recent study of concussions in the NFL, Pellman et al. (58, 59) recommended a value below 250 to minimize the risk of Mild Traumatic Brain Injury (MTBI or concussion).

Holbourn (33) worked with gel models of the brain and showed that rotational acceleration was an important mechanism in head injury. Ommaya and Hirsch (55) scaled primate head injury data to humans and predicted that a level of head rotational acceleration in excess of 1,800 r/s^2 would have a 50% probability of cerebral concussion in man. Analysis of injuries produced in rhesus monkey experiments resulted in Gennarelli et al. (23) estimating a 16,000 r/s^2 rotational acceleration tolerance threshold in man.

In a recent survey of rotation head injury criteria, Ommaya et al. (54) found that the rotational acceleration of 4,500 r/s^2 was required to produce concussion in an adult and that severe diffuse axonal injuries (DAI) occurred at 18,000 r/s^2. The range is from scaling of animal impact data and indicates the difficulty in developing a precise injury-prediction criterion for rotational motion, since the shape and mass of the animal brains are different from human and scaling laws assume geometric similarity. This makes extrapolating animal data to humans difficult. Also, the low mass of the animal brain requires very high rotational accelerations to produce closed head injuries (55). The combination of these factors makes predicting injuries in humans difficult. Furthermore, Pellman et al. (58, 59) found concussion was related to translational acceleration of the head.

In an effort to understand the relationship between forces delivered in boxing and risk of head injury from linear and rotational accelerations, the biomechanics of four different types of boxing punches was studied. The punches included straight punches to the forehead and jaw, a hook and an uppercut. Olympic boxers threw punches at an instrumented Hybrid III headform with their dominant hand, except for the hook. Correlations were made between the biomechanics of the Hybrid III head responses for the boxing punch and helmet impacts in professional football with attention to concussion. This study shows the similarities and differences between the head impacts.

Methodology

Eleven Olympic boxers weighing 51 kg (112 lb) to 130 kg (285 lbs) were included in the study. They were tested while participating in the 2004 United States Boxing National Championships. The research received approval from the Wayne State University's Human Investigation Committee and each boxer read and signed an informed consent prior to testing. Since their involvement was voluntary, they could withdraw from the study at any time. A certified boxing trainer was present for the tests even though the risk of injury was minimal. The boxers were not compensated for their participation.

Each boxer was evaluated for four punches. After a boxer warmed-up, they were instructed to strike the instrumented Hybrid III head with their gloved fist two times with four different punches, straight punches to the forehead and jaw and a hook and uppercut. Based on the method of Walilko et al. (97), three of the four punches were delivered with the dominant hand, including a straight punch to the jaw, a straight punch to the forehead and an upper cut to the jaw. For the fourth punch, the boxers' non-dominant hand was instrumented and they were asked to deliver a hook to the temple. Impact location of the punch was determined by high-speed video.

Measurement of Effective Hand-Arm Mass

Height and weight of each boxer were measured and anthropometric data for the dominant hand was collected. Volume measurements were obtained by submerging the dominant fist up to the styloid process in a water bath (97). The displaced volume of water was measured. Boxers then submerged their fist and forearm up to the epicondyles of the humerus bone. The effective mass of the fist and forearm were calculated from the anthropometric and volumetric measurements by estimating the density of the body and converting densities of the forearm and hand. The segmental forearm and hand densities were multiplied by the segmental volumes to determine mass. Density estimates (d) were made using $d = 0.6905 + 0.0297c$ with $c = h/w^{1/3}$, where h = height (inches), w = weight (lbs), conversion factor for hand density (1.08) and conversion factor for forearm density (1.06). The equations have been shown to be suitable for estimating segmental body masses (8, 10, 100).

Test Setup

The test methods follow those described by Walilko et al. (97). A Hybrid III dummy with a frangible face was used to represent the response of the jaw and realistically transfer acceleration to the head. For the tests, a cork insert was used to give facial compliance for the straight jaw punches. The straight blow to the forehead and hook to the temple were on regions of the Hybrid III with known biofidelity. The uppercut was to the jaw, which has less biofidelity.

The tests used the compliant face of the Hybrid III head (50). This design has an improved biomechanical response in the facial region over the standard molded Hybrid III and more accurately reproduces the force and acceleration of the head for impacts in the frontal, zygomatic, maxillary and mandibular regions. Other devices used either a stiff load-measuring face or deformable structures in regions other than the jaw (1, 52, 92). The head and neck of the dummy were attached to the upper torso, which was fixed to the table by the flexible lumbar joint. This gave realistic head and upper body motion.

For the straight punch to the forehead, hook and uppercut, the punches were directly to the Hybrid III head. The straight punch to the jaw loaded the compliant face. Headgear was not placed on the dummy. The upper torso was attached to a rigid table with a foam pad placed below the Hybrid III abdomen insert so that the dummy remained in an upright position after each punch. Scaffolding was used to adjust the height of each boxer to a minimum 175 cm (69″) to ensure the punches were in the horizontal plane.

The Hybrid III simulates a tensed neck so the head is normally upright. The segmented neck includes flexible polymer discs to simulate the flexion-extension and lateral bending responses. A cable inside tightens the assembly to give the right neck response in calibration testing and during head acceleration (77, 80). While the Hybrid III neck was utilized in the study, it is unknown how it represents the strength of a boxer's neck. Boxers undergo extensive training to develop the neck muscles necessary to resist punch forces from an opponent. However, Johnson et al. (35) demonstrated that neck muscle tension has little effect on the oscillation of the head under sinusoidal excitation from a shaker.

Instrumentation

Instrumentation was placed in the boxer's clenched hand. Two Endevco (San Juan Capistrano, CA) 7264-2k accelerometers were secured to a semicircular cylinder, which was wrapped with the boxer's hand to measure hand acceleration in a biaxial arrangement. Integration of acceleration gave the velocity change of the hand during the punch. The Hybrid III was equipped with the standard triaxial accelerometers (Endevco 7264-2k) at the head center of gravity (cg) and six more accelerometers in a ″3-2-2-2 configuration″ to determine rotational acceleration (56). Processing of the nine accelerations determined the complete three-dimensional motion of the head. Rotational accelerations were computed from linear accelerations in the head. The analysis is valid for accelerometers coincident with the origin of the system or coincident with one of the axes. Deviations from this were required in the Hybrid III head, and a correction for centripetal and Coriolis acceleration was made according to DiMasi (16). Data was collected at 10,000 Hz using the TDAS PRO (DTS, Inc.) data acquisition system (SoMat, Co, Urbana, IL) and post processed according to SAE J211-1 (78).

Video Film Analysis and Target Location

A target was placed on the glove to digitize its motion and calculate impact velocity. Additional targets were attached to the head of the Hybrid III to measure the overall kinematics of the dummy during impact. Images were captured with a Kodak HG2000 high-speed video camera. The camera recorded the event at 4500 images per second. Digitization of the data was completed using the Image Express for video recording and processed according to SAE J211-2 (79).

Data Collection Procedure

After an appropriate warm-up period, the boxer was asked to lightly punch the head of the instrumented dummy with their wrapped and gloved hand. If there was no pain or discomfort, they were asked to increase their punch strength until they reached a point where they were throwing ″normal″ punches. Once the boxer was comfortable throwing punches, they were asked to deliver four different punch types to the dummy. Each punch type was performed twice for a total of eight punches per boxer. The order of punch placement was varied randomly; however, all of the punches for a particular hand were completed before the alternate hand was tested.

Punch and Head Inertial Forces

Impact forces were determined by two methods. First, the hand acceleration was measured for each punch and multiplied by their effective punch mass, which was determined separately. This estimated the impact force for the punch. Second, the resultant head acceleration of the Hybrid III was multiplied by the head mass of 4.45 kg to estimate the inertial force on the head. The punch force includes the inertial force on the head and neck loads, so it was always higher than the inertial force. The severity of the impacts was further quantified using translational and rotational acceleration, head injury criterion (HIC), severity index (SI) and change in head velocity (ΔV).

Concussion Risks from Boxing Punches

Pellman et al. (58) determined concussion risks using the Logist function in the Statistical Analysis Package (SAS). This function relates the probability of concussion p(x) to a response parameter x based on a statistical fit to the sigmoidal function $p(x) = [1+\exp(\alpha-\beta x)]^{-1}$, where α and β are parameters fit to the NFL response experience from the laboratory reconstruction of game impacts. The risk of concussion was determined for Olympic boxer punches using the NFL risk functions based on all football players exposed to helmet collisions. The parameters for the Logist functions were $\alpha = 2.677$ and $\beta = 0.0111$ for HIC, $\alpha = 4.678$ and $\beta = 0.0573$ for translational acceleration and $\alpha = 5.231$ and $\beta = 0.000915$ for rotational acceleration.

FE Modeling of Brain Responses

Head accelerations from the Hybrid III dummy were used as input to a finite element (FE) model of the boxer's brain. This analysis follows the approach reported by Viano et al. (95) in the study of brain responses in NFL concussions. The brain responses for three punches from the heaviest boxer were simulated and compared with the patterns of brain deformation determined with NFL concussions. Early and mid-late strain responses and brain displacement were determined to show timing and areas of greatest brain deformation from the punches.

Effective Impact Radius

During a punch, the head experiences translational and rotational acceleration from the impact force. The accelerations are coupled. The impact can be resolved into a force at the head cg and a moment. The moment is related to the punch force times a radius between the impact axis and head cg. The radius (r) of impact causing rotational acceleration can be approximated by a simplified 2D relationship: $r = \alpha I/F = (\alpha/a)(I/m)$, where the head mass (m) is 4.45 kg and moment of inertia (I) about a lateral axis through the head cg is 0.022 kg m^2 (45). The impact radius is proportional to the ratio of rotational to translational acceleration, where the constant of proportionality is the ratio of head moment of inertia to mass, or 0.0049 m^2. If the simplified analysis assumes the head and neck are acting together to resist the impact, the mass is 5.80 kg and the average moment of inertia is 0.035 kg m^2 (4). This gives a ratio of head-neck moment of inertia to mass of 0.0059 m^2 and the effective radius increases 20%.

RESULTS

Boxer Anthropometry and Effective Punch Mass

Table 1 shows anthropometric data on each boxer. It also gives the volume of the hand and forearm, which were used to determine the effective punch mass. The average weight of the 11 boxers was 76.5 ± 22.1 kg (167.7 ± 48.5 lb) and their height was 177.2 ± 9.2 cm (69.8 ± 3.6 inch). The hand mass increased with boxer weight and averaged 1.67 ± 0.28 kg.

Punch Force and Biomechanical Responses

Figure 1 shows an example of the punch kinematics for the four different blows. These are images from the high-speed video and show the progression of the punch to the head of the Hybrid III dummy. The left column shows the straight punch to the jaw, which initially causes flexion of the upper neck and extension of the lower cervical region (third image). This happens because the punch is below the head cg. The

TABLE 1. Anthropometry of the Olympic boxers[a]

Boxer #	Height (cm)	Weight (kg)	Hand volume (mL)	Forearm volume (mL)	Forearm circumference (cm)	Wrist circumference (cm)	Wrist width (cm)	Wrist thickness (cm)	Wrist to knuckle (cm)	Fist width (cm)	Fist thickness (cm)	Hand mass (kg)
2	165	50.9	320	1130	26.2	16.2	5.3	3.8	10.2	8.5	6.7	1.26
11	168	55.0	350	1270	26.2	16.1	5.6	3.7	10.7	8.3	6.4	1.41
13	168	57.7	350	1260	27.1	16.5	5.5	3.7	9.7	8.5	6.1	1.39
4	180	65.9	360	1300	27.5	17.0	6.0	4.0	8.0	8.5	6.5	1.45
9	173	70.9	470	1730	29.5	17.0	5.4	4.5	10.9	8.5	6.9	1.88
10	178	70.9	460	1660	27.8	16.5	5.7	4.2	12.2	8.9	7.0	1.83
14	175	74.5	490	1840	30.4	18.5	6.3	4.7	9.8	9.4	7.5	2.00
8	177	84.1	440	1540	30.0	18.0	5.7	4.4	10.2	9.1	6.8	1.66
7	188	87.3	540	1520	31.2	18.7	6.4	4.6	12.7	9.5	7.5	1.68
12	183	91.8	440	1590	30.8	18.5	6.1	4.6	11.8	9.2	6.7	1.72
6	196	129.5	470	2030	32.7	19.0	6.0	4.3	10.8	9.0	7.0	2.15
Average	177.2	76.2	426	1534	29.0	17.5	5.8	4.2	10.6	8.9	6.8	1.67
SD	9.2	22.1	70	276	2.2	1.1	0.4	0.4	1.3	0.4	0.4	0.28

[a] SD, standard deviation.

FIGURE 1. *Sequences from high-speed video of a boxer throwing a straight punch to the jaw* (left column*) and forehead* (second column*), a hook* (third column*) and an uppercut* (right column*). Time is shown in the top left of each image and the number in the top right is the frame count.*

hook produces lateral bending of the neck and twists the head about its vertical axis. The uppercut is shown in the last sequence at the right.

Table 2 summarizes the average biomechanical responses for the four different punch types. The full data is given in the *Appendix*. The hook produced the greatest impact force (4405 ± 2318 N) and inertial load on the head (3107 ± 1404) with an average neck load of 855 ± 537 N. It also had the highest change in hand velocity (11.0 ± 3.4 m/s). The lowest forces occurred with the uppercut to the jaw. Because of the range in boxer weight, the overlap in responses does not produce significant differences when the punches are grouped by type.

The hook also produced the largest head translational and rotational accelerations, reaching an average 71.2 ± 32.2 g and 9306 ± 4485 r/s^2, respectively. The straight punch to the jaw resulted in the largest neck loads (1088 ± 381 N) as the neck flexes as the jaw is driven rearward. This also resulted in the greatest bending moment of 81.9 ± 23.8 Nm about the y-axis (flexion-extension bending).

Even though the hand velocity change was in the range of 6.7 to 11.0 m/s on average, the change in velocity of the Hybrid III head was only 2.8 to 3.1 m/s on average for the various punches. This reflects the relatively low punch mass during the momentum exchange in the punch, although the change in hand velocity includes some effects of rebound from the punch.

Comparing Boxer Punches to NFL Concussions

Figure 2 shows the average and standard deviation in head inertial force (head mass times acceleration) for the four punches from the Olympic boxers and three conditions from NFL helmet impacts. The highest force is for NFL players experiencing concussion. Lower forces were measured in the NFL reconstructions for players struck without injury and for the striking players in helmet-to-helmet tackles. The force from the boxer's hook exceeded that of the non-injured NFL players and was within the statistical range for concussion. The jaw and forehead impact forces were lower and the uppercut produced the lowest inertial loads on the Hybrid III head.

Figure 3 compares the head biomechanical responses for the NFL game reconstructions and the boxer punches to the Hy-

TABLE 2. Summary punch forces and biomechanical responses of the Hybrid III[a]

Punch type	HIC15	SI	Res. head acc. g	Head delta V m/s	Res. rot. acc. r/s^2	Res. rot. vel. r/s	Res. neck load N	X-neck moment Nm	Y-neck moment Nm	Z-neck moment Nm	Res. hand acc. g	Hand delta V m/s	Peak force using head N	Punch force using hand N
Forehead														
Average	58	72	47.8	3.1	5452	22.9	664	−8.0	39.6	4.8	206.7	8.2	2085	3419
SD	44	53	20.1	0.7	2107	5.9	199	8.9	26.6	3.0	75.2	1.5	876	1381
Hook														
Average	79	99	71.2	3.1	9306	29.3	855	34.6	8.4	−14.8	263.4	11.0	3107	4405
SD	70	87	32.2	1.0	4485	6.2	537	21.1	6.6	8.1	105.4	3.4	1404	2318
Jaw														
Average	52	66	48.8	2.9	6896	20.7	1088	−21.1	81.9	10.3	145.8	9.2	2127	2349
SD	42	53	20.9	1.0	2848	5.6	381	22.2	23.8	6.3	57.1	1.7	910	962
Uppercut														
Average	17	23	24.1	2.8	3181	17.5	1486	−12.0	−21.0	6.5	92.9	6.7	1051	1546
SD	19	25	12.5	0.9	1343	5.0	910	5.2	6.1	3.4	39.9	1.5	547	857

[a] HIC15, head injury criterion for 15 ms duration; SI, severity index; Res, resultant; Acc, acceleration; Rot, rotation; SD, standard deviation; Vel, velocity.

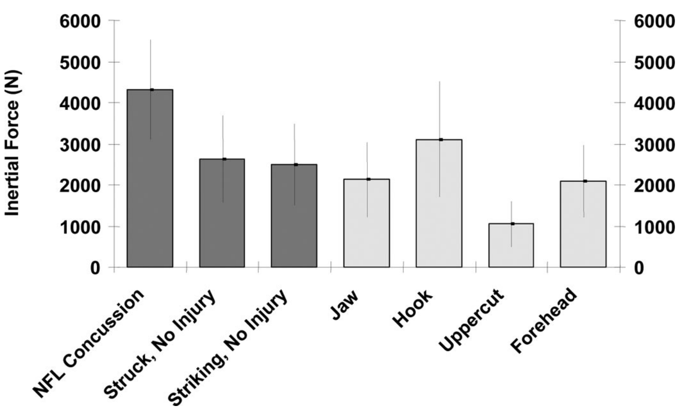

FIGURE 2. *Inertial force on the Hybrid III head for NFL game impacts and four different boxing punches.*

brid III dummy. The HIC was higher for NFL players experiencing concussion. The NFL players not injured or striking without injury had HICs slightly higher than the range of boxer punches. A similar trend can be seen in the translational acceleration, but the hook shows levels in the range for a risk of concussion based on the NFL experience. Interestingly, the boxers deliver more rotational acceleration to the Hybrid III dummy head for the hook and jaw punches than occurred in NFL concussions. However, the duration of impact is shorter for the boxing punch, so the rotational velocity of the head is similar to that in NFL concussion impacts with longer duration but lower rotational acceleration.

Figure 4 shows the peak rotational and translational accelerations for the NFL concussions and the boxers punches to the Hybrid III dummy. The closed circles represent concussed players in the NFL and the open symbols biomechanical data from players struck without injury or the striking players. The boxer data is included showing that jaw and hook punches have impacts in the range of the concussions experienced in the NFL. On average the boxers produce more rotational than translational acceleration with their punches.

Concussion Risks in Boxing and NFL Head Impacts

Table 3 shows the average and standard deviation in concussion risk using biomechanical responses from the Hybrid III dummy. Risk functions were used for concussion, where HIC had the strongest statistical correlation with NFL concussions (58). Based on HIC, the hook had a 13.8% ± 14.3% risk of concussion. On average, the predicted risks were in the range of 7 to 14% for the various punches. Similar concussion risks were predicted by peak translational acceleration, which was also a good predictor of concussion for the NFL impacts. The risk averaged 11 to 97% based on the peak rotational acceleration.

For comparison, the average and standard deviation in NFL concussion risk is shown from Pellman et al. (58, 59). Based on 28 players struck with 22 concussions, the average risk of concussion was 58.2% ± 33.0% based on HIC. While no striking player experienced concussions, the head responses were high enough to estimate a risk of 23.4 ± 20.7%, which obviously overstates the incidence based on the field experience. Nonetheless, the Logist risk functions were determined as a

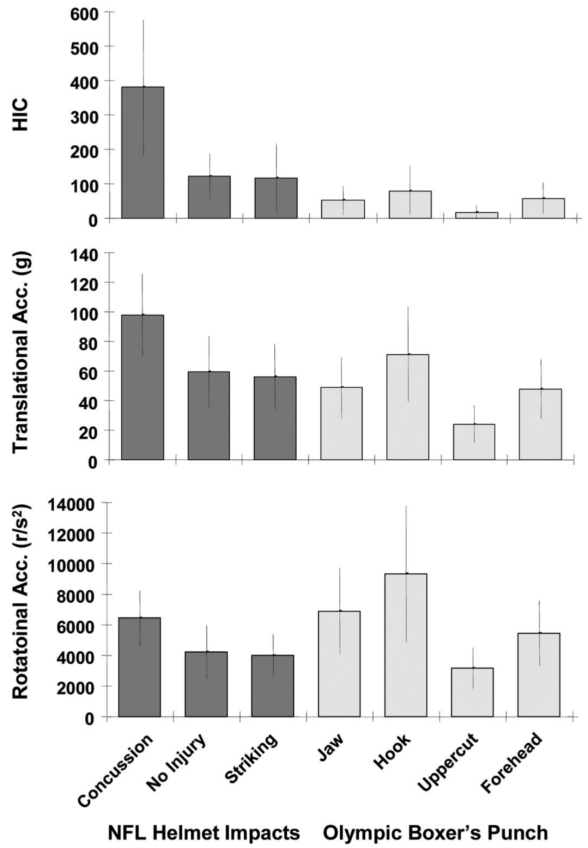

FIGURE 3. *HIC and peak translational and rotational acceleration for NFL game impacts and four different boxing punches.*

probability relationship between measured biomechanical responses and physician observed concussions.

FE Modeling of Brain Responses

Figure 5 shows the simulated brain responses for a hook from the heaviest Olympic boxer (boxer #6). This was a substantial blow and the strain "hot spots" show an early pattern in the temporal lobes with the highest strain occurring late in the midbrain. The maximum strain occurred about 10 milliseconds after the peak impact force.

Effective Impact Radius

Figure 6 shows groupings of peak accelerations for NFL concussions, striking players without concussion, struck players without injury and the four boxer punches. The lines are based on the relationship: $r = \alpha I/F = (\alpha/a)(I/m)$. The average radius between the punch axis and head cg was 57 to 71 mm with 57 mm for the forehead punch, 65 mm for the hook, 66 mm for the uppercut and 71 mm for the jaw punch. In contrast, NFL concussions occur at a higher translational acceleration and force on the head, but lower rotational acceleration. The average radius varied from 32 to 36 mm for the NFL

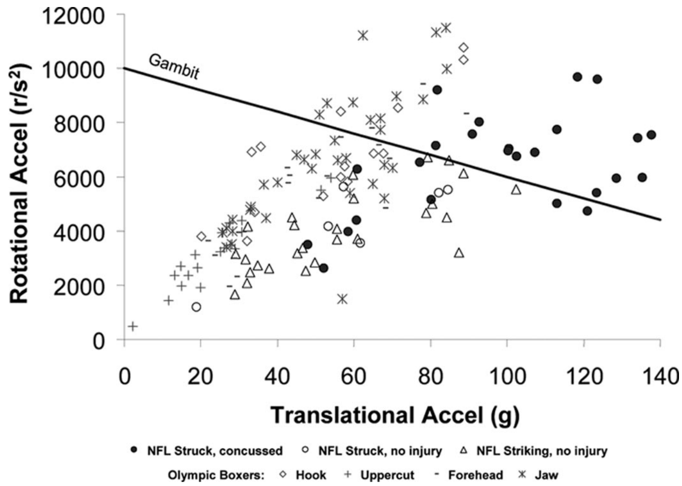

FIGURE 4. *Individual data points for translational and rotational acceleration of the Hybrid III head for NFL game impacts and four different boxing punches.*

players struck and injured (35 mm), those struck and not injured (36 mm) and the striking players (32 mm).

The ratio of head rotational to translational acceleration is higher in boxing than in NFL helmet impacts and results in a larger effective radius. The effective radius in football is 48% smaller than in boxing on average (34 mm v 65 mm). If the simplified analysis assumed the head and neck was acting together to resist the impact, the effective radius increases 20%.

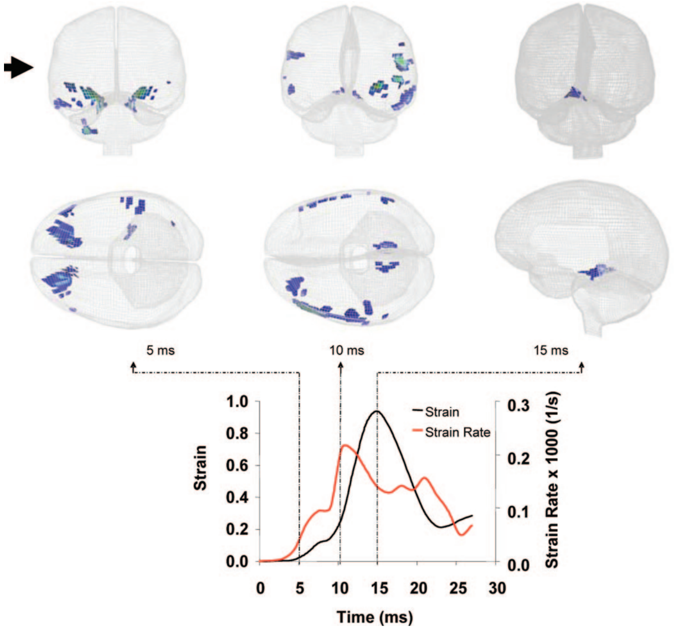

FIGURE 5. *Strain "hot spots" in the brain for a hook (Test 6h) showing the early, mid and late response pattern. The punch is to the right side of the Hybrid III head. The peak translational acceleration occurred at 5 milliseconds and the duration was about 8 milliseconds. The strain and strain-rate responses are shown for tissue in the brain.*

TABLE 3. Estimated risk of concussion for boxer punches based on the National Football League concussion experience[a]

		Risk of concussion		
		HIC	Trans acc. g	Rot. acc r/s^2
Olympic boxer punches	Forehead			
	Average	11.2%	11.9%	49.2%
	SD	5.7%	16.5%	32.5%
	Hook			
	Average	13.8%	35.9%	96.9%
	SD	14.3%	31.9%	29.3%
	Jaw			
	Average	10.5%	12.5%	78.0%
	Standard Deviation	5.2%	18.1%	35.2%
	Uppercut			
	Average	7.3%	3.1%	11.1%
	SD	1.8%	4.1%	15.9%
NFL Helmet Impacts	Struck players			
	Average	58.2%	57.7%	61.8%
	SD	33.0%	29.9%	28.8%
	Striking players			
	Average	23.4%	24.6%	26.0%
	SD	20.7%	22.7%	22.1%
	All players			
	Average	40.8%	41.2%	43.9%
	SD	32.5%	31.2%	31.2%

[a] HIC, head injury criterion; Trans acc, translational acceleration; Rot acc, rotational acceleration; SD, standard deviation.

DISCUSSION

This study compares the biomechanical forces affecting the head and brain from boxing punches with football helmeted impacts occurring in the NFL. There were three significant differences noted. The boxers' punches resulted in lower translational accelerations in the struck head, as compared to the football impacts. The boxers' punches applied a higher moment to the struck head than did the football impacts. This necessarily resulted in higher rotational accelerations in the head struck by the boxers' punch. Boxers therefore sustain brain injury by two mechanisms, translational and rotational accelerations of the brain, with a preponderance of the rotational component. Professional football players, on the other hand, sustained MTBI mostly by translational accelerations. These differences in the biomechanical forces may help explain the clinical differences between head injury in boxing and professional football.

Brain injury resulting in death is mostly due to acute subdural hematomas, which is much more common in boxing than in professional football. This difference can be explained by the differing effects on the bridging veins and other fragile brain structures resulting from the biomechanical forces. In addition, boxers are susceptible to a specific, unique pattern of chronic brain damage that has never been seen in American football players. The relative preponderance of rotational accelerations and the lower translational accelerations seen in boxing impacts may set the stage for boxers to sustain this type of long term brain injury.

This study also demonstrates one significant similarity between the head impacts in boxing and professional football. In both cases, the FE modeling indicates that the highest strain and strain-rate occur in the midbrain in the late time frame after the peak head acceleration. This suggests that high midbrain strain in the late timeframe might be a final common pathway in the development of concussion from a variety of head impact conditions.

Difference Between Boxing and NFL Brain Injuries

Acute head injuries in boxing can be more serious and devastating than those seen in professional football. There were over three hundred boxing deaths recorded in England due to brain injury before 1937 (66). Between 1945 and 1980, there were over 335 documented fatalities due to boxing (38–40, 66, 86–89). There have been many more deaths due to boxing in the years since 1980.

Approximately 75% of brain injury deaths from boxing are due to subdural hematoma (40, 66, 75, 89). Most of the other 25% are due to other traumatic intracerebral hemorrhages. Most acute subdural hematomas that account for the majority of boxing deaths are due to tearing and rupture of the bridging veins that run between the dura and the surface of the brain (40, 66, 89). These fragile structures are easily torn by head trauma. Studies in animals as well as clinical experience in humans indicate that almost all of the subdural hematomas are due to the effects of rotational forces stretching the bridging veins (89). One would expect that tearing of the bridging veins and subsequent subdural hematoma would be seen more commonly after head blows in a sport such as boxing with a preponderance of rotational acceleration as compared to professional football in which translational forces predominate.

The data from these tests on boxers shows proportionately higher rotational accelerations than translational acceleration in boxing. There is another possible explanation for tearing of the bridging veins in boxing. The present study indicates that

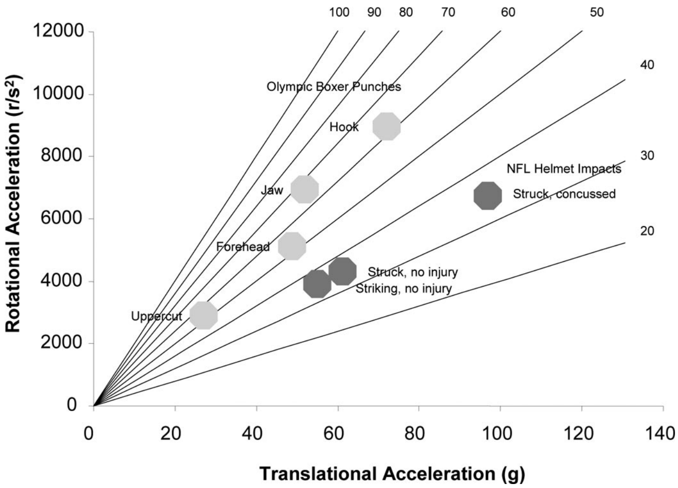

FIGURE 6. *Peak translational and rotational acceleration for Olympic boxing punches and NFL helmet impacts. The lines are a constant distance between the axis of impact force and head center of gravity (cg) using a simplified formula linking translational and rotational acceleration.*

punches directly to the jaw cause flexion of the head and neck resulting in stretching (high strains) of the bridging veins. In contrast, punches impacting the forehead cause extension of the head and neck with resultant compression (low strains) of the bridging veins.

Although there have been 433 fatalities due to head injury in football between 1945 and 1984 (337 cases due to subdural hematoma), almost all of these have occurred in high school or college players (66), where neck musculature and abilities are not as well developed as in the professional athlete. The authors are aware of one head injury related death in a professional Canadian football player and no brain injury related deaths in American professional football players since 1945. There was one case of subdural hematoma occurring in an American professional football player during those years but this was successfully removed without fatality.

The occurrence of subdural hematoma is consistent with the relatively larger translational accelerations and inertial force on the head in professional football than in boxing. This also raises the possibility that rotational head accelerations may be more predominant among college and high school players than in professional football players. It is also possible that the younger high school and college athletes' brains are more susceptible to tearing of the bridging veins than the more mature adult brains of the professional players. Although the absolute number of head injury related fatalities are similar in boxing and amateur levels of football, the incidence is in fact much higher in boxing because of the much larger number of annual participants in football at all levels compared to the number of annual participants in boxing.

Another clinical difference between the patterns of brain injury seen in boxing and football is the chronic brain damage seen in boxers but not in football players. A chronic encephalopathy of boxers has been well known to physicians since its initial description in 1928 (39, 47, 49, 75, 89). The clinical syndrome of pyramidal, extra pyramidal and cerebellar dysfunction combined with organic mental syndromes with cognitive and memory impairments and personality changes has been well documented (75, 89). The chronic encephalopathy may range from very mild to very severe.

During the past 25 years, studies have documented a pattern of cognitive and memory impairments in boxers ranging from subclinical to clinical dementia of varying degrees from chronic brain injury (5, 76). Studies have also defined a specific pattern of neuropathology which constitutes the chronic encephalopathy of boxers. This consists of abnormalities of the septum pellucidum, the cerebellum, the substanta nigra and the cerebral hemispheres (11, 89). The abnormalities of the septum pellucidum region include tears and fenestrations with resultant CSF leakage into the septum resulting in cavum septum pellucidum. The cerebellar findings consist of scarring and loss of Purkinje cells. In the substantia nigra there is depigmentation and loss of neurons. There is cerebral scarring as well as the presence of neurofibrillary tangles without senile plaques. There is enlargement of the third and lateral ventricles. This distinct neuropathological pattern is diagnostic of chronic encephalopathy of boxers. Studies have demonstrated that this chronic encephalopathy of boxers is related to the accumulation of multiple subconcussive blows to the brain over a long period of time. Its occurrence is directly related to the length of a boxer's career and the number of bouts fought, not to the number of times the boxer had been knocked out (5, 75, 76).

Roberts' (75) study suggested that chronic encephalopathy was more prevalent in heavier weight class boxers. This certainly would be consistent with the findings of the present study that heavier fighters delivered blows with higher forces than those generated by the lighter boxers. Critchley's (14) earlier paper, however, found that chronic encephalopathy occurred equally across all weight classes thus raising a cautionary note to interpreting the findings of this present study. This syndrome has never been reported in American football players. The present study may give some insight into why this syndrome is seen in boxers but not in other athletes such as professional football players. The present results indicate that boxers' brains sustain translational forces which are largely at or below the threshold for MTBI in NFL players and are likely to be subconcussive in nature when encountering an alert opponent rather than the stationary Hybrid III dummy. As a result, boxers are infrequently knocked out and thus able to continue fighting even though there may be substantial force in the punches landed.

In the course of training or a bout, a boxer may sustain large numbers of such blows in a repetitive manner. Repetitive head impacts with relatively high translational and rotational acceleration sustained over a period of time may cause tearing of structures such as the septum pellucidum resulting in cavum septum pellucidum, damage to the deep midline structures of the brain such as the substantia nigra and damage to the cerebellum and cerebral hemispheres. The nature of the forces impacting the boxer's brain may ultimately make him susceptible to long term chronic brain damage (39, 71). The professional football player's head, on the other hand, is occasionally subject to much higher translational accelerations which are more likely to result in cerebral concussion and more likely to result in the player being removed from play or at least limited in his game activities for at least a short period of time. Professional football players do not sustain frequent repetitive blows to the brain on a regular basis. In addition, the relative preponderance of translational forces in professional football players may make them less susceptible to chronic injury than does the relative preponderance of rotational accelerations in boxing.

Difference Between the Biomechanics of Boxing and NFL Head Impacts

Figure 2 shows the inertial force on the head (58, 59) from reconstruction of helmet impacts in the NFL. They collected data from Hybrid III dummies simulating impacts recorded on game video. The laboratory reconstructions provided data on the biomechanical responses associated with recorded concussions in the players, and other severe impacts without injury. No concussion occurred in six struck players, and none of the striking players was injured. Interestingly, the boxers in this study generated impact forces that are similar to the non-concussion forces on the helmeted heads of NFL players.

The super-heavy weight boxer generated inertial forces of 3633 ± 1196 N and punch forces of 5352 ± 2775 N, which is at the average impact force causing concussion in NFL players. Since a majority of NFL players are injured by facemask or lateral impacts on the helmet, the loading direction is consistent with directions of the boxing punches, although straight anteroposterior impacts are an uncommon cause for NFL concussions. Boxer weight correlated with punch force, HIC and head acceleration in tests on Olympic boxers (97). While weight was a good predictor, punch force had a stronger correlation with HIC and translational acceleration. This means the effective mass of the boxer's punch is more important in increasing the severity of a blow.

There are probably two means by which boxers deliver concussive blows. The first means involves the boxer delivering enough translational acceleration. The hook involves a blow to the temple, which is just above the head cg. The forehead punch delivers force frontally above the head cg and the jaw impact applies force below the head cg. These impacts translate the head cg, and the forces can reach levels consistent with NFL concussions. The damaging mechanism is translational acceleration where the greater the mass of the punch, the greater the head HIC and translational acceleration. The second means involves rotational acceleration, which occurs with the impacts taking advantage of the offset from the head cg. During the punch, the axis of impact moves away from the head cg and introduces proportionately more rotational acceleration during the punch. The hook, for example, is always thrown with the elbow bent (43). This necessarily results in the axis of impact moving away from the cg after impact, thus imparting a significant amount of rotational acceleration to the opponent's head. The rotational nature of the hook has led sports writers to describe this punch as "whirling" and "tornadic" (43).

Increasing the effective mass behind a punch is the best way to increase the force of the punch. Increased acceleration of the punch may also result in increased force but it would seem difficult to increase the acceleration in the actual scenario of throwing a punch. Boxers and trainers intuitively realize that increasing the effective mass behind the punch increases its force. Champion boxer Jack Dempsey wrote that when he threw a straight left hand punch he began his attack "with a falling step forward toward the target with his left foot" (43). This "started the weight transfer which was the power source." He continued, "as you take your falling step forward, you shoot a half open left hand straight along the power line chin high" (43). The emphasis on keeping the arm and hand straight is consistent with the results from Walilko et al. (97) indicating that keeping the wrist straight and not flexed increases the force of the punch. By falling forward into the punch, Dempsey was increasing the effective mass behind the punch and thus increasing its force. Dempsey also knew that it was very difficult to deliver a knockout punch without throwing the entire weight of his body behind the punch. He wrote that it would be very difficult to throw a knockout punch by just turning the shoulders (43).

Presumably, other champion fighters and trainers have learned the same lessons that were known to Dempsey (96). The results of the present study lend scientific validity to this intuitive knowledge.

Figure 3 shows the HIC and translational and rotational accelerations from Pellman et al. (58, 59) on the biomechanics of NFL concussion. The boxers cause HIC and peak translational accelerations in the lower range of concussions in the NFL, but the head rotational accelerations can be higher from boxing punches. This means the boxers do not transfer as much energy in their punch as the collisions in the NFL. *Figure 4* shows more of the trend in the peak translational and rotational acceleration. In this case, there is a greater overlap in the peak rotational accelerations of the boxing impacts with concussion levels found for NFL players. However, the peak translational accelerations are lower than what occurs in the NFL. GAMBIT is a head injury criterion that limits the combination of rotational and translational acceleration (53). The tolerance line is shown.

With the use of football helmets, the striking player must line up his impacts closely with the head cg of the other player. This allows the impact to transfer energy. If the impact vector is at an angle, the blow will glance off due to the smooth plastic shell of the helmets. Players realize that they need to align their impact through the head cg to deliver a solid blow and maximize energy transfer to the other player. Severe helmet impacts that cause concussion involve high translational acceleration and change in head velocity (ΔV). NFL concussions involve an average impact velocity of 9.3 ± 1.9 m/s; and, the ΔV is 7.2 ± 1.8 m/s for the concussed player. Since the duration of impact is nominally 15 milliseconds, the peak head acceleration is high at 98 ± 28 g. In football, there is a strong correlation between translational and rotational acceleration due to the impact alignment and subsequent head-helmet motion.

In boxing, the punch and glove conform more to the head of the opponent allowing punches to induce high rotational acceleration without high translational acceleration. The effective mass of the boxer's fist is 1.67 ± 0.28 kg, which is more than an order of magnitude lower than the 25 kg effective mass of the helmeted football player who strikes an opponent (64, 93). With concussion, the striking player lines up their head, neck and torso so their effective mass is considerable, and only the head and part of the opponent's neck resist the blow. In boxing, the most efficient energy transfer involves more rotational acceleration than translational acceleration.

The punch velocity of the boxers averaged 6.7 to 11.0 m/s for the four different punches. These levels are essentially similar to the impact speed in football concussions; but, the head ΔV after a punch was only 2.8 to 3.1 m/s on average, well below half that with NFL concussion. This reflects the much lower effective mass of the punch. Boxers cannot deliver high translational acceleration and ΔV to the opponent because of the low punch mass in comparison. Obviously, the effectiveness of punches is greater when the opponent is dazed and their neck muscles are more relaxed, since this lowers the effective head mass resisting the punch. Many of the well known boxing fatalities in the modern era have involved a fighter who has been dazed and stunned by multiple blows from his opponent. He is in a defenseless state with resultant marked diminution of muscle tone in the cervical paraspinal muscles (89). The resultant decrease in effective head mass results in increased translational and rotational accelerations of the head with every further punch. These increased accelerations are more likely to result in high strains to the brain, including the bridging veins, leading to severe injury or death.

While this discussion is theoretical, it is based on the mechanics of two different sports that can deliver neurocognitive effects to the brain in the form of memory, cognitive and functional problems. What is critical to the logic is that striking players in the NFL do not experience concussion even though they have head ΔV of 4.0 ± 1.2 m/s and the same impact velocity as the concussed player. Their ΔV and peak translational acceleration are above what the boxer can deliver in their punches. This indicates that rotational accelerations may be a factor in boxing knockouts, since translational effects are low.

Obviously, boxers can deliver rotational accelerations in and above the range where NFL players are concussed. However in both sports, we have not determined the root cause of concussion or knockouts. The underlying injury mechanism may depend on yet unknown combinations of translational and rotational acceleration, or factors of the brain response to skull accelerations associated with the impact. It is clear that head accelerations displace the skull in a complex kinematic, which loads the brain and causes internal stresses that deform neural tissues (45, 93). Brain and spinal cord tissues are sensitive to the rate and extent of strain in an impact (91). A sufficient combination of strain and strain rate can bruise the tissue and cause dysfunctions in neural function.

FE modeling of the brain response to a hook is shown in *Figure 5*. The responses are similar to the patterns of strain "hot spots" found in NFL concussions (95). Strain migrates from the temporal lobes early in the response to the midbrain, where the largest strain occurs late after the primary impact force of the punch. Since most of the NFL concussions involve lateral acceleration of the head, the hook has a similar direction of head loading. The boxers claim the hook is their knockout blow, but it is hard to deliver in a bout. Obviously, more analysis of the FE responses is needed to determine strategies for improving head protection in boxing and football; but, these results offer new insights into the biomechanical responses of the brain during head impact.

Effective Impact Radius

Figure 6 shows typical NFL concussion conditions and those of the striking and struck players who were not injured. The average radius was 34 mm for the three groups (range 32-36 mm). The average radius for the four boxer punches was 65 mm (range 57–71 mm). The lines are constant radii. The impact radius is proportional to the ratio of rotational to translational acceleration with the ratio of moment of inertia to mass a constant of proportionality. The ratio of head rotational to translational acceleration is larger in boxing than in NFL impacts. This simplified analysis seems to point to rotational acceleration as a possible factor in the severity of knockout punches, whereas the NFL concussion studies found the strongest correlation with translational acceleration and that the impacts had to be aligned with the head cg to prevent the helmets from sliding off. The duration of impact is shorter for boxing punch.

This analysis points to two different biomechanics of head injury in boxing. One associated with high translational acceleration and HIC, and another related to high rotational acceleration with low translational acceleration and HIC. Obviously, more study is needed to determine the underlying causes of boxing knockouts and football concussions. The simplified analysis assumed average values for a complex three dimension event. Also, the radius varies with time, the punch can be at varying orientation to the head cg and the flexibility of the neck is a factor. Nonetheless, the simplified analysis shows that a punch produces proportionately larger rotational than translational accelerations than in football by having a larger effective radius. The analysis also shows that rotational acceleration depends on the translational acceleration; they are inextricably coupled in an impact.

The data generated by this present study has been compared to the data from MTBI in the National Football League. The present study indicates that boxer punches cause relatively lower translational accelerations to the Hybrid III dummy head than the impacts seen in the NFL; whereas, the rotational accelerations are similar or higher to those seen in NFL concussions. This indicates that there may be a greater role for rotational acceleration in boxing blows compared to translational accelerations in helmet impacts in professional football. These results suggest that rotational acceleration of the head may be a factor in chronic brain damage from repetitive impacts in boxing, since the pattern of brain injury is not seen in football players.

Comparing Concussion Risks in Boxing with NFL Head Impacts

Previous studies have primarily determined the force of a boxer's punch using a heavy bag or instrumented pendulum. The current study is a continuation of the effort started by Walilko et al. (97) to collect head impact responses using the Hybrid III dummy to determine head injury risks from boxer punches. The head-neck assembly of the Hybrid III closely represents the mass and compliance of the human head and neck. With this system, the risk of injury in terms of HIC, and translational and rotational acceleration can be explored from the momentum transfer of a punch to the head.

Because the dummy has humanlike impact responses and there are risk functions for concussion, the results of this study are relevant to determining the concussion risks from the punches of boxers. Although the knockout punch is a dramatic part of the sport, it is a relatively uncommon event. Studies have shown that knockouts occur in less than 5% of professional fights and probably in less than 2 to 3% of amateur level fights and in less than 1% of all amateur fights (40, 48).

The present results are consistent with this data. The translational accelerations resulting from these punches were at levels that one would expect to see relatively few clinical concussions, particularly for the uppercut and forehead impacts. The rotational accelerations are relatively higher and perhaps closer to the levels that one might see with clinical concussion. Most of the rotational accelerations were at or above levels expected to cause cerebral concussion in football. The uppercut is the exception. Therefore the laboratory results are consistent with a measurable but low incidence of knockout in the sport of boxing.

High rotational accelerations were found. The average peak rotational acceleration varied from a high of 9308 rad/s^2 for the hook to 3181 to 6898 rad/s^2 for the other punches. Ommaya et al. (54) indicated a rotational acceleration of approximately 4500 rad/s^2 were required to produce concussion. He also stated severe DAI occurs at 18,000 rad/s^2, and moderate and mild DAI occur at 15,500 r/s^2 and 12,500 r/s^2, respectively. Earlier studies by Pincemaille et al. (65) measured rotational accelerations of 13,600 rad/s^2 and rotational velocities of 48 rad/s during boxing. There were no cases of concussion in the tests. The current tests with the Hybrid III also show high rotational accelerations, however, the data reflect higher tolerances than specified in the literature otherwise knockouts would be much more common in boxing matches. Since the rotational acceleration tolerances are based on scaling of animal data, a question may be raised about the adequacy of the technique, which assumes similar geometry and equivalent material characteristics between animal and man.

Using risk functions for concussion in NFL players (58), the boxer impacts in this study show the highest risks with peak rotational acceleration (*Table 3*). The average risk was 11 to 97% for the four boxing punches. However, the NFL data showed the most significant correlation of concussion with HIC and peak head acceleration. Based on those biomechanical responses, the boxing data indicate a concussion risk of 7 to 14% for HIC and 3 to 36% for translational acceleration. Interestingly, the boxers do not generate enough head ΔV to reach much of a concussion risk based on the NFL data. HICs were low for the four punches with a risk of severe traumatic brain injury $<2\%$ (70). The average HICs were 17 to 79 also well below the proposed NFL concussion threshold of 250 (58, 59).

Limitations

Force delivered to the jaw loads the dummy in an area with responses that are similar to the human, so the reported translational acceleration and HIC reflect what occurs in boxing (50). However, it is uncertain how force from the uppercut to the jaw is related to risk of MTBI or if the Hybrid III dummy, in its current form, has sufficient similarity to the human response to measure risks with this punch. Development of human surrogate with an articulating jaw may improve the response of the head in this region and may show different biomechanical responses for the uppercut.

Before applying the results of this study to actual boxing experience in the ring, one must be aware of other limitations. The boxers that participated in this study were Olympic level amateur boxers. Although these boxers are at the higher echelon of amateur boxers, they most likely have not attained the proficiency or power levels of professional boxers. It is probable that more accomplished professional boxers can deliver punches with significantly higher force than those that were generated by these boxers.

It also must be pointed out that only four specific punches were evaluated in this study, including a straight punch to the jaw, uppercut to the jaw, hook to the temple and forehead punch. Other punches and uppercuts, hooks and crosses to other regions of the head may have different characteristics than the punches studied here; and, therefore may result in different translational and rotational accelerations in the opponent's brain than were seen in this study.

Furthermore, the data in this study was collected in a controlled laboratory setting, not during an actual boxing match. Factors such as fatigue, excitement and the effects of "adrenaline" on the boxers may significantly alter the forces of the punches delivered. Also, in an actual boxing match, the movements and defensive maneuvers of the opponent may affect the forces of the punches delivered to the opponent's head. In the present study, the punches were delivered to a stationary dummy head. The punch forces that were measured may be different than those that are seen in an actual boxing match.

APPENDIX: Measured and determined results from testing of Olympic boxers[a]

Boxer No. punch ID punch type	HIC15	SI	Res. head acc. g	Head delta V m/s	Res. rot. acc. r/s ^{2}	Res. rot. vol. r/s	Res. neck load N	X-neck moment Nm	Y-neck moment Nm	Z-neck moment Nm	Res. hand acc. g	Hand delta V m/s	Peak force using head N	Punch force using hand N
Forehead														
2f	15	18	23.1	2.5	3062	24.1	453	−18.3	29.5	3.0	87.4	8.4	1006	1082
11c	66	86	55.9	3.4	7448	21.6	825	−23.2	71.0	9.4	219.4	9.0	2438	3037
11f	87	106	66.0	3.4	7159	21.9	711	−10.8	41.5	7.9	298.7	9.4	2878	4134
13b	56	69	52.5	2.9	6017	10.6	789	−28.8	79.5	2.6	234.2	9.1	2286	3198
13e	82	101	67.8	3.4	4835	20.7	1105	−3.0	11.4	5.0	239.6	10.0	2955	3272
4c	9	13	21.3	2.1	3622	19.7	561	−2.3	60.1	10.2	165.1	8.2	930	2350
4f	16	23	29.2	2.5	3938	28.1	491	−1.9	40.3	7.0	159.7	7.2	1273	2273
9c	16	19	26.8	2.1	1928	16.5	376	−1.8	11.7	2.7	91.0	6.1	1171	1682
9f	18	22	28.9	2.3	2313	16.8	349	−1.3	13.7	3.2	90.7	6.2	1259	1676
10e	47	57	50.2	2.6	5202	20.3	701	−1.3	53.0	7.1	184.5	9.2	2187	3306
10h	53	61	42.7	3.1	6060	20.1	570	−9.6	38.6	4.6	221.4	7.0	1861	3968
14c	55	69	42.2	3.9	5764	28.6	683	−22.0	55.8	—	309.1	8.3	1842	6049
8c	15	19	28.2	2.1	2971	21.1	515	−0.7	7.0	0.6	87.4	5.0	1230	1422
7d	121	148	68.7	4.3	6663	25.0	920	−1.3	18.2	2.4	263.0	9.1	2994	4321
7g	123	145	64.2	4.1	7793	38.9	747	−10.4	53.3	3.4	273.2	9.1	2801	4488
12d	26	35	32.2	3.0	4747	23.0	556	−1.8	47.6	3.2	284.4	6.3	1405	4784
12g	31	42	42.1	2.8	6327	19.8	946	−9.1	98.9	10.0	290.4	9.4	1837	4886
6b	120	150	77.5	3.9	9421	30.1	585	−2.2	11.1	2.6	207.8	9.8	3379	4376
6e	145	181	88.9	3.9	8307	25.8	739	−1.9	10.4	1.7	221.3	9.9	3875	4661
Average	58	72	47.8	3.1	5452	22.9	664	−8.0	39.6	4.8	206.7	8.2	2085	3419
SD	44	53	20.1	0.7	2107	5.9	199	8.9	26.6	3.0	75.2	1.5	876	1381
Hook														
2a	57	70	57.6	3.2	6386	27.1	579	20.2	7.7	−5.7	181.6	10.7	2513	2247
2b	72	87	65.1	3.5	6862	27.8	780	19.6	5.6	−6.2	275.4	11.7	2839	3407
11g	111	142	92.2	3.7	12562	19.4	1081	54.3	13.5	−20.9	294.7	11.3	4022	4079
11h	77	96	71.4	3.2	8536	28.8	641	26.9	5.6	−13.6	310.1	10.6	3114	4292
13g	58	69	58.4	3.0	5804	24.3	448	13.4	3.2	—	293.0	9.3	2545	4001
13h	52	66	56.6	2.7	5967	22.1	446	8.6	5.7	−9.2	212.6	8.5	2470	2904
4a	74	90	67.7	3.3	6863	28.8	476	23.5	4.2	−4.6	313.7	14.0	2952	4466
4b	102	142	126.6	3.8	17487	36.0	1763	22.1	3.3	−6.2	281.5	19.8	5519	4007
9g	20	25	32.1	2.1	3635	21.5	491	18.8	1.9	−9.6	111.6	7.2	1400	2063
9h	55	64	52.0	2.8	5282	24.9	550	14.5	3.3	−6.0	272.2	10.7	2267	5030
10a	16	19	34.1	2.1	4693	27.9	494	30.7	3.6	−18.1	214.4	8.5	1488	3843
10b	11	14	33.2	1.6	6907	46.3	709	56.8	18.0	−27.1	55.4	5.9	1447	994
14a	87	116	102.3	3.5	12083	22.5	36	98.4	31.7	−27.8	270.9	12.6	4462	5302
14b	107	137	114.7	3.1	13337	37.0	1929	49.8	11.3	−21.4	443.0	12.4	5001	8671
8i	5	8	20.1	1.4	3804	36.6	315	21.2	6.2	−24.8	110.0	6.4	875	1791
8j	54	72	88.6	2.2	10756	35.2	840	27.8	8.4	−27.5	301.5	9.7	3862	4907
7a	96	118	92.9	3.6	19925	26.9	2167	43.6	7.9	−9.1	296.1	10.2	4052	4865
7b	95	116	80.4	3.5	12381	32.0	1304	62.3	9.5	−22.2	197.2	12.6	3508	3241
12a	19	26	35.6	2.5	7117	28.8	975	56.3	6.0	−13.7	200.2	9.2	1552	3368
12b	52	67	56.6	2.7	8396	32.0	652	34.6	11.1	−18.6	231.0	9.0	2470	3887
6g	187	221	88.6	4.6	10301	29.4	950	29.6	4.7	−8.3	472.4	14.4	3863	9950
6h	330	415	140.6	5.7	15626	28.4	1178	27.4	13.2	−9.0	455.7	17.7	6131	9597
Average	79	99	71.2	3.1	9305	29.3	856	34.6	8.4	−14.8	263.4	11.0	3107	4405
SD	70	87	32.2	1.0	4485	6.2	537	21.1	6.6	8.1	105.4	3.4	1404	2318

APPENDIX: Continued

Boxer No. punch ID punch type	HIC15	SI	Res. head acc. g	Head delta V m/s	Res. rot. acc. r/s ^{2}	Res. rot. vol. r/s	Res. neck load N	X-neck moment Nm	Y-neck moment Nm	Z-neck moment Nm	Res. hand acc. g	Hand delta V m/s	Peak force using head N	Punch force using hand N
Jaw														
2e	11	14	28.3	1.5	4422	19.6	842.3	−22.8	46.9	12.5	82.0	10.2	1232	1015
2h	20	25	36.5	2.2	5706	11.0	1020.4	−12.3	68.3	10.0	134.1	8.7	1590	1659
11d	49	64	53.0	3.0	8706	26.0	966.2	−21.1	97.4	16.9	190.1	10.3	2311	2632
13c	66	82	64.3	3.5	8097	16.8	954.4	−5.4	95.5	10.0	227.9	9.7	2804	3113
13f	66	81	55.8	3.1	6605	14.4	812.1	−23.2	80.7	—	250.8	10.3	2435	3424
4e	15	29	32.9	2.1	4775	19.0	899.2	−14.8	60.1	12.6	100.9	7.7	1433	1436
4h	15	21	28.3	2.3	3979	22.4	768.0	−9.4	73.5	15.4	124.4	8.0	1236	1771
9a	12	16	25.8	2.2	3946	18.6	706.7	−24.5	63.4	5.8	54.7	7.2	1124	1010
9e	15	21	26.8	2.5	4108	19.8	823.8	−15.2	77.0	6.3	68.2	6.5	1168	1260
10d	76	95	62.2	3.9	11215	25.8	2101.2	−104.3	50.7	26.8	113.1	9.2	2715	2028
10g	10	13	26.6	2.0	3399	13.5	664.0	−6.5	58.6	2.5	95.2	7.1	1159	1707
8a	77	100	55.0	4.4	7334	23.9	1350.5	−25.7	107.6	8.1	95.4	7.0	2399	1553
7e	120	153	84.1	3.8	11492	28.8	1711.6	−10.8	117.8	11.5	203.3	11.3	3666	3340
7h	80	103	59.7	4.3	8738	28.2	1453.6	−34.2	116.6	15.3	175.0	10.6	2603	2876
12c	12	16	28.1	1.7	3511	11.3	906.8	−12.1	59.1	9.9	174.2	8.6	1224	2930
12f	32	40	45.0	2.0	6805	24.5	1082.0	−18.7	96.7	8.8	172.4	12.0	1961	2901
6a	130	168	81.5	4.1	11321	25.5	1427.7	−3.8	120.1	2.4	168.2	10.9	3553	3543
6d	121	149	84.2	3.4	9969	24.6	1101.4	−14.1	83.7	0.8	194.2	11.0	3671	4090
Average	52	66	48.8	2.9	6896	20.7	1088.4	−21.1	81.9	10.3	145.8	9.2	2127	2349
SD	42	53	20.9	1.0	2848	5.6	381.4	22.2	23.8	6.3	57.1	1.7	910	962
Uppercut														
2d	19	24	28.8	2.8	3338	24.4	1496.0	−18.5	−18.5	10.3	96.1	8.9	1257	1189
2g	18	24	25.5	3.6	3946	22.0	1729.2	−8.2	−25.6	5.7	96.9	8.8	1113	1198
11b	6	0	2.1	2.8	472	5.5	73.6	−1.9	−4.0	1.6	33.1	4.6	92	458
11e	9	14	19.1	3.5	2632	15.8	1424.9	−9.3	−27.6	5.7	55.8	5.7	831	773
13a	12	18	26.1	2.8	3381	20.3	1865.2	−12.2	−27.8	4.8	87.6	6.7	1140	1197
13d	20	27	30.7	2.9	4387	10.0	2175.7	−20.3	−15.1	5.8	98.4	8.5	1339	1343
4d	10	14	18.6	3.1	3125	22.6	1291.9	−10.1	−28.3	8.3	48.2	6.7	809	686
4g	3	4	14.9	3.6	1973	14.8	819.0	−6.1	−22.5	4.9	70.2	5.9	651	999
9b	4	5	13.1	1.9	2359	15.8	919.6	−13.0	−22.7	4.4	63.1	4.4	572	1167
9d	8	9	19.9	1.7	1919	12.7	909.9	−8.7	−16.0	6.8	107.3	6.4	867	1983
10c	4	7	14.7	2.1	2707	15.4	815.6	−15.8	−26.3	11.8	67.0	5.1	640	1201
10f	12	17	25.1	2.4	3218	18.2	1383.6	−13.9	−25.6	7.3	112.3	5.5	1096	2012
8b	23	30	27.7	3.4	3478	21.6	1499.2	−14.1	−21.6	12.6	74.7	6.1	1209	1216
7c	5	7	16.6	1.7	2354	19.8	850.9	−5.3	−16.1	2.1	74.4	7.5	725	1223
7f	19	27	30.6	3.3	3964	17.2	2044.5	−19.3	−22.2	7.7	78.5	8.0	1337	1291
12e	2	3	11.6	1.1	1431	15.0	419.8	−11.1	−17.0	8.3	95.1	4.2	507	1600
12h	14	20	27.5	3.0	4288	15.7	1347.0	−20.3	−15.1	11.8	192.9	7.2	1200	3246
6c	73	94	53.9	4.2	5950	20.2	3658.7	−11.3	−21.9	2.9	156.2	8.0	2350	3270
6f	64	85	51.3	4.2	5515	25.5	3500.2	−9.3	−25.6	1.4	158.1	8.5	2238	3330
Average	17	23	24.1	2.8	3181	17.5	1485.6	−12.0	−21.0	6.5	92.9	6.7	1051	1546
SD	19	25	12.5	0.9	1343	5.0	910.5	5.2	6.1	3.4	39.9	1.5	547	857

[a] HIC15, head injury criterion for 15 ms duration; SI, severity index; g, gravity; Res acc, resultant acceleration; Res rot acc, resultant rotational acceleration; N, newton; SD, standard deviation.

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COMMENTS

Viano et al. continue their fascinating biomechanical studies of sports-related traumatic brain injury by investigating boxing punches and comparing them to helmet impacts in professional football. Rotational acceleration of the head is proportionately greater after boxing punches, whereas translational acceleration tends to predominate in football impacts. As the authors suggest, this fundamental biomechanical difference may account for the higher incidence of both acute injuries, such as acute subdural hematoma, and chronic brain injury in boxing.

Alex B. Valadka
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Viano et al. carefully studied the biomechanics of boxing injuries using finite element modeling, and compared the physical characteristics of the impact to the head from an amateur boxer's punch with the physical characteristics of head injuries sustained in professional football. As they point out in the discussion section, the clinical applicability of their results is limited. For example, the forces applied to the head as measured in their laboratory would be expected to cause a much higher incidence of concussion than is actually seen on the playing field. It also is likely that professional boxers sustain significantly greater translational and rotational displacement of the head than do amateur boxers. Because of the more severe rotational forces sustained by boxers compared with football players, and the much higher incidence of lethal brain injuries in boxing, I completely agree with the statement by Viano et al. that "boxers are susceptible to a specific, unique pattern of chronic brain damage that has never been seen in American football players." Their study underscores the need for studies that clearly define the histopathology of chronic traumatic encephalopathy in professional football players, and cautions against generalizing the autopsy findings in boxers to other sports.

Donald W. Marion
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In sports, such as American football, ice hockey, boxing, rugby, lacrosse, and martial arts, in which contact is an integral part of the game, athletes develop hitting strategies to gain an advantage. In many cases, this strategy involves head impacts. This article provides insight into what, until now, has largely been ignored. Depending on the situation, athletes are able to use different strategies to create concussions. Even though it is obvious that a number of factors contribute to head injuries in sport, helmet designs have, for the most part, focused on preventing subarachnoid bleeds. This has primarily been accomplished by managing a 40 to 80 joule impact under 275 gs of linear acceleration. Recently, cerebral concussion has become more of a concern, whereas subarachnoid bleeds have become relatively rare. Although the threshold for protecting the brain against concussive injuries is not well established, it has been estimated at approximately 4500 r/s^2 for angular acceleration and 75 to 80 gs of linear acceleration. Football helmets are primarily designed to manage impacts to prevent subarachnoid bleeds and although they do provide some protection against concussion, they are not designed to do so.

The objective of these and many other sports is to manage physical interactions in such a way as to enrich the competitive environment of the activity without undue risk of injury to participants. This thin line is managed by game rules, coaching and training programs, protective equipment, and player integrity. A breakdown in any one of these elements increases the risk of player injury. Coaches, players, and game officials all have incentives to allow increased hitting. Examples of this are numerous and include coaches having to gain an advantage over other teams, especially when the team is under pressure to win, players fighting to win a spot on the team, and league officials under pressure to increase fan attendance. Professional sports are extremely vulnerable to the pressures of producing a product that is attractive to a broad audience.

This article demonstrates the need for research to better understand the mechanism underlying head injuries in specific activities. There is little doubt athletes become extremely skilled in gaining an advantage and, if necessary, can use any number of strategies to take advantage of their opponents. Just how athletes receive concussions in sport is still not well understood. The recent National Football League study identified linear acceleration as the primary mechanism for concussions in professional football. This included a very limited data set and should be interpreted accordingly. More extensive research directed at understanding the mechanism of head injuries in individual sports would be extremely valuable in managing head injuries in all sports.

T. Blaine Hoshizaki
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In a further contribution from the National Football League (NFL) Mild Traumatic Brain Injury Committee, the authors sought to evaluate the potential relationships between the impact biomechanics of boxing and football. To do so, 11 Olympic-level boxers (weight range 112–285 lbs) delivered 78 blows (inclusive of four types including hooks, uppercuts, and jabs) to the head of a Hybrid III Dummy. Recorded variables included translational and rotational head acceleration and neck load. Punch force was measured using a biaxial acceleraton model. The most significant change in hand velocity was in the hook (11.0 m/s) in addition to the greatest punch force (4405 ± 2318 N). The authors found that boxing punches have proportionately more rotational than translational acceleration than what is observed in football cuncussions.

This is useful information and indirectly addressed a number of questions. It is interesting to note that the NFL Mild Traumatic Brain Injury Committee has yet to identify an example of dementia pugilistica (or variants thereof) in retired NFL players. Obviously, this is quite the contrary in professional level boxing. Exposure obviously differs significantly both with regard to the quantity of head impacts and the likely associated force in the absence of headgear.

The importance of head-gear will need to be defined at the amateur level before any consideration at the professional level. Even then, the more spurious elements involved in the management and promotion of the sport will likely be slow in adopting any suggested guidelines regarding headgear use at the professional level. The perception is the possible impact on the fan base and the associated income. Initial work continues at the Olympic Training Facility in Colorado to evaluate responses in the presence and absence of headgear in boxing. Evaluation of impact at varying distance will also be required.

The consideration of headgear is important at the professional level. A lack of understanding of the components of injury that are most significantly related to central nervous system injury and short- or long-term disability remain unknown. Conceptually, the understanding of the force associated with "Heavy Hands," or lack thereof, in athletes susceptible to concussion (Glass Jaw) will also need to be determined. An important first step would be to evaluate the potential myth of the big punch. Obviously, the only chance an individual may have in a fight in which, based upon points, the individual has no chance of winning the fight. The damage sustained by individuals who are losing in these competitions tends to be consistent. We would suspect that the suggestion of an early end to a 12-round championship fight given an inability to win based on the 10 point scale would be as well received as the use of head gear. A final caveat. There are a number of things about the sport of professional boxing that we cannot and should not change.

> *'I am going to kill him good. I do not care about styles. Styles do not mean anything. I have seen every style in the world. I have been in this game for 18 years. I have been a world champ for 12. He cannot even touch that. I am going to be the WBA heavyweight champ of the world. I am ready to go no matter what. I do not care: you want to play rough, I will play rough. Boy, you have no idea. I tell you, I will hurt you.'*

Min Park
James Toney
John Arthur
Michale L. Levy
San Diego, California

This study by Viano et al. investigates the biomechanical forces involved in boxing punches in comparison with football impacts. Using a Hybrid III test dummy and accelerometers, 11 Olympic boxers were studied delivering hooks, uppercuts, and straight punches to the instrumented Hybrid III test mannequin. Using finite element analysis, the kinetic energy and head responses were compared with similarly determined impacts from video documented and reconstructed concussions in NFL players. Any study that analyzes the biomechanical forces imparted during contact sports is a welcome addition to the literature. As the authors note, closed head injury is an occupational hazard of many sports, particularly in boxing and football, in which neurocognitive effects to the brain can occur in the domains of memory, cognitive, and functional injury. The ability to carefully visualize and measure the various aspects of the contact athlete's torso and head response to high velocity impacts increases our understanding of the issues involved.

This study also adds insight and helps clarify the fact that translational and rotational accelerations are major components of the human body's response in contact sports. In my opinion, the effective radius of the impact and the relative contribution of the supporting musculature in the thorax, head and neck are most important factors.

Our computerized video analysis of various types of boxing matches has led to the conclusion that there is little difference between the fights which are considered "classic" versus those that result in a lethal outcome. Unquestionably, the most significant contributing factor in lethal boxing outcomes is the absorption of multiple blows to the cranium, especially during long-duration fights and with a fighter who is progressively impaired, resulting in relaxation of the supporting neck musculature and lowers the effective head mass against the punch.

Surprisingly, the dramatic single-punch knockout rarely results in a lethal outcome or significant brain injury in a boxer. The chronic effect seen in boxers is related to an accumulation of blows to the cranium, accentuated with years in the sport and in heavier weight classifications. the authors note the limitations of their study, which was performed in a controlled laboratory setting with Olympic-level amateur boxers, looking at a limited number of punches, and those delivered to a stationary test mannequin. This is an important study with comparative data between two contact sports which are associated with potential for brain injury.

Julian E. Bailes
Morgantown, West Virginia

Hoard unearthed from a bog in Norway containing a gold trefoil brooch from France, a large golden neckring from Russia, and coins from Arabia, Byzantium, and England. (Courtesy of Museum of Cultural Heritage, Oslo).