An Investigation of the NOCSAE Linear Impactor Test Method Based on In Vivo Measures of Head Impact Acceleration in American Football

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The performance characteristics of football helmets are currently evaluated by simulating head impacts in the laboratory using a linear drop test method. To encourage development of helmets designed to protect against concussion, the National Operating Committee for Standards in Athletic Equipment recently proposed a new headgear testing methodology with the goal of more closely simulating in vivo head impacts. This proposed test methodology involves an impactor striking a helmeted headform, which is attached to a nonrigid neck. The purpose of the present study was to compare headform accelerations recorded according to the current (n = 30) and proposed (n = 54) laboratory test methodologies to head accelerations recorded in the field during play. In-helmet systems of six single-axis accelerometers were worn by the Dartmouth College men's football team during the 2005 and 2006 seasons (n = 20,733 impacts; 40 players). The impulse response characteristics of a subset of laboratory test impacts (n = 27) were compared with the impulse response characteristics of a matched sample of in vivo head accelerations (n = 24). Second- and third-order underdamped, conventional, continuous-time process models were developed for each impact. These models were used to characterize the linear head/headform accelerations for each impact based on frequency domain parameters. Headform linear accelerations generated according to the proposed test method were less similar to in vivo head accelerations than headform accelerations generated by the current linear drop test method. The nonrigid neck currently utilized was not developed to simulate sport-related direct head impacts and appears to be a source of the discrepancy between frequency characteristics of in vivo and laboratory head/headform accelerations. In vivo impacts occurred 37% more frequently on helmet regions, which are tested in the proposed standard than on helmet regions tested currently. This increase was largely due to the addition of the facemask test location. For the proposed standard, impactor velocities as high as 10.5 m/s were needed to simulate the highest energy impacts recorded in vivo. The knowledge gained from this study may provide the basis for improving sports headgear test apparatuses with regard to mimicking in vivo linear head accelerations. Specifically, increasing the stiffness of the neck is recommended. In addition, this study may provide a basis for selecting appropriate test impact energies for the standard performance specification to accompany the proposed standard linear impactor test method. [DOI: 10.1115/1.4000249]

1 Background

The incidence and severity of mild traumatic brain injury (mTBI) or concussion in sports has recently come under increased scrutiny by both researchers and the public, particularly in helmeted sports including American football and hockey. Leagues, coaches, medical staff, families, and players are interested in ways to reduce the incidence of mTBI without changing the spirit of the hard-hitting game. The advent of helmet test standards developed by the National Operating Committee for Standards in Athletic Equipment (NOCSAE) in the early 1970s and the implementation of these standards by major sports governing bodies helped to significantly reduce the incidence of traumatic brain injuries and deaths, particularly in football [1]. However, these test methods were not explicitly developed with the goal of reducing mTBI.

The NOCSAE standard drop test method [2] was developed in response to the increasing number of head and cervical spine injuries occurring during American football play. Specifically, the advent of the facemasks in 1954 led to a noticeable increase in the number of head and cervical spine injuries: With their faces protected, players began to intentionally block and tackle with their heads. During the following 15 years, the number of head and cervical spine fatalities occurring during organized football play quadrupled, peaking with 36 fatalities in 1968 [1]. The neurological and biomechanical literatures of the era, which facilitated the development of the NOCSAE standard for football helmets, focused primarily on extended loss of consciousness, coma, and fatal head trauma in experimental canines and skull fracture in human cadavers [3–8]. Laboratory test methods were developed to simulate the conditions measured by these experimental models. Following the implementation of the NOCSAE helmet certi-

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[2]Conflict of Interest: Authors R. G., J. C., and J. C. have a financial interest in HIT System technology used to record in-vivo head impacts for this study.

Contributed by the Bioengineering Division of ASME for publication in the Journal of Biomechanical Engineering. Manuscript received October 9, 2008; final manuscript received August 18, 2009; accepted manuscript posted September 18, 2009; published online December 17, 2009. Assoc. Editor: Michael Sacks.

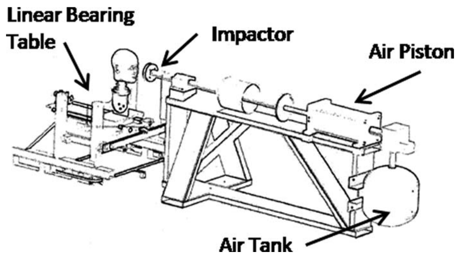

Fig. 1 Sketch of linear impactor test apparatus [15]

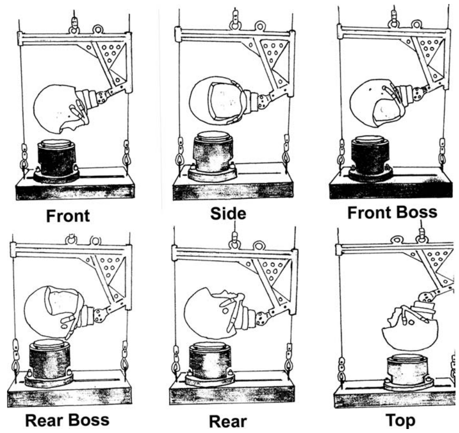

Fig. 2 The NOCSAE linear drop test setup for all required impact locations [28]

fication standard in 1973 and a series of rule changes preventing head first tackling in 1976, the annual number of fatalities from football related head and cervical spine injuries was reduced to less than 10 by 1980 [1].

In the four decades since the NOCSAE standard was established the body of scientific knowledge regarding the mechanisms, incidence, and long-term effects of mTBI has grown substantially. Langlois et al., [9] estimated that the incidence of sport and recreation related mTBI in the United States was between 1.6×10^6 and 3.8×10^6 annually. In a recent study of collegiate football players, nearly 5% of all athletes sustained a concussion in a single season and 15% of those injured athletes sustained a repeat concussion in the same season [10]. Concussions in sports make up only 7% of all mTBI cases presenting to emergency departments in the United States. However, sport is the leading cause of mTBI for the 5–14 age group, representing 27% of all cases [11]. The National Institutes of Health (NIH) have declared mTBI to be a serious public health issue and reducing the incidence, severity, and postinjury symptomology of mTBI has become a national research priority [12,13]. The increased research, treatment, and prevention efforts are further justified by the health care costs associated with mTBI in sports, which are estimated to be in the hundreds of millions of dollars annually [14].

NOCSAE has developed but not yet implemented a new method for standardized headgear testing, the standard linear impactor (LI) test method [15]. The stated goal of the proposed LI test method (referred to as proposed NOCSAE standard) is to "more closely emulate on-field impacts believed to be responsible for mTBI" (see page 1 of Ref. [15]). This proposed methodology involves a projected impactor striking a helmeted headform attached to a Hybrid III neck (Denton ATD, Rochester Hills, MI). The head/neck unit is attached to a linear slide table aligned with the direction of the impact that allows the headform and the impactor to move in the direction of impact after initial contact (Fig. 1). The NOCSAE headform is an anthropomorphic head model used in both the current and proposed NOCSAE standards. It was developed to match mechanical impedance and geometrical properties of human cadaver skulls [8]. This head model is an alternative to metallic half-sphere headforms, which are utilized in many sports helmet test standards. The NOCSAE headform consists of a blow molded polyethylene inner skull filled with glycerin to simulate the brain and overmolded with a polyurethane compound. The assembly also contains a hollow sinus cavity and mounting hardware encapsulated in the area of the foramen magnum. The full head assembly is overmolded with a polyurethane skin that includes facial features.

The proposed NOCSAE standard test method differs dramatically from the current NOCSAE test method [2], which prescribes a twin rail linear drop test setup consisting of the NOCSAE headform mounted on a rigid aluminum drop bar propelled by gravity onto a 1.27 cm (0.5 in.) thick firm (38 durometer, shore A) rubber pad (Fig. 2) [16]. The proposed NOCSAE standard requires that a facemask be affixed for all test impacts, including an impact directly to the center of the facemask, unlike the current methodology in which helmets are tested without facemasks. The prescribed impact locations for the new protocol are faceguard, front boss, side, rear boss, and rear. Additionally, the "testable region" or locations on the helmet that may be selected for the random test impact were modified to reflect the inclusion of facemask impacts: The random impact location is selected by the tester, is not the same as any of the prescribed impact locations, and must be within the testable region on the helmet (Fig. 3). Due to the configuration of the proposed test apparatus it is not currently possible to align the neck and the impactor, thus preventing a pure crown test impact. As such, the LI test apparatus in its current form supplements but does not replace the current drop test apparatus.

It is prudent to test helmets by impacting them in similar ways and in similar locations as they will be impacted during field use. In other words, a helmet test standard should mimic the dynamics of head impacts occurring during play. Specifically, it is desirable for a helmet test to mimic the frequency properties of in vivo head impacts because the energy attenuation properties of most materials, including foams and elastomers used in sports helmets, are frequency dependent.

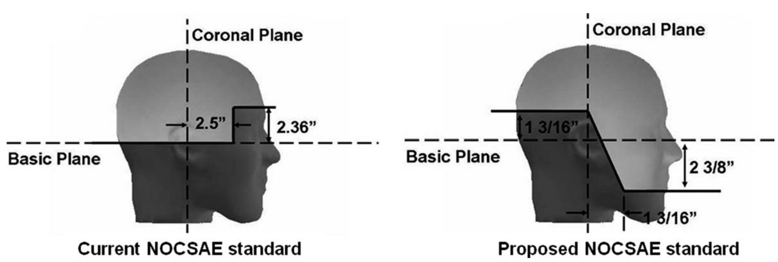

Fig. 3 Testable locations (shown in light gray) for the current and proposed NOCSAE standards

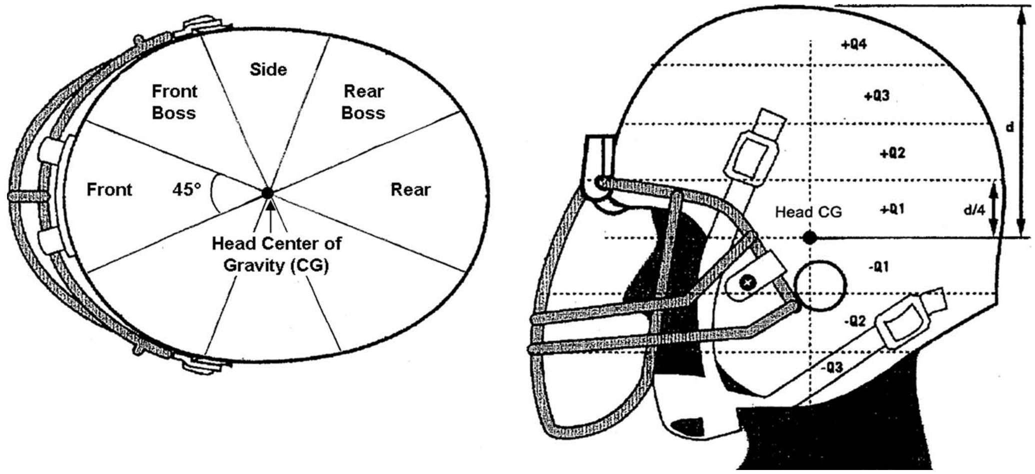

Fig. 4 Impact location bins based on azimuth (left) and elevation (right) [30]

To assess whether the recently proposed NOCSAE standard achieved its aim of more closely simulating in vivo head impacts, a method to monitor and record head accelerations resulting from in vivo head impacts was required. The head impact telemetry (HIT) System (Simbex, Lebanon, New Hampshire; Riddell, Elyria, Ohio) was utilized to monitor and record head impact accelerations during all games and practices. The HIT System is a telemetry based data acquisition system that records the linear acceleration of the head during impact using an in-helmet unit (IHU) consisting of six single-axis accelerometers [17]. The results of in vivo head acceleration measurements using HIT System technology have been reported by the authors and others [18–27]. This study has focused on quantifying the frequency and severity of impacts sustained by collegiate and high school football and hockey players.

The purpose of the present study was to utilize football head impact data recorded in vivo to evaluate the current and proposed NOCSAE football helmet testing protocols and to provide a basis for the selection of appropriate impactor velocities for the standard performance specification to accompany the proposed standard test methodology. There were three goals, as follows: (1) to compare frequency of the in vivo impact locations with the locations specified in the current and proposed standard test protocols; (2) to determine the LI velocities that are required to produce headform acceleration measures similar to the highest acceleration measures recorded in vivo; and (3) to compare frequency properties of linear head accelerations recorded in vivo to frequency properties of linear headform accelerations generated by the current drop test protocol and by the proposed LI protocol.

2 Methods

2.1 In Vivo Data Collection Methods. In vivo impact data were collected from instrumented helmets worn by 40 football players ($n=20{,}733$ impacts) at Dartmouth College (Hanover, NH) during practices and games for the 2005 and 2006 seasons. Helmets were Riddell Revolution (Riddell, Elyria, OH) instrumented with HIT system (Simbex, Lebanon, NH) IHUs containing six single-axis accelerometers. Each impact recorded was automatically time-stamped, associated with an anonymous player identification number, and stored in a local database, which was later synchronized with a central server. Only impacts with a resultant peak linear acceleration >10 g were considered for analysis (as in Refs. [23–25]). The participating players were selected by the medical staff from the pool of volunteers at each school such that a wide variety of positions were covered. Starters were given preference over second-team players. Players selected their helmets as per school protocols. Informed consent was obtained for all participating athletes in accordance with the Dartmouth College Institutional Review Board.

Measured data included impact location and resultant center-of-gravity (CG) head linear acceleration recorded over a 40 ms window [17] from which Gadd severity index (GSI) (Eq. (1)) was computed [6]. GSI is the integral of linear head acceleration (exponentially weighted) versus time

$$\mathrm{GSI} = \int a^{2.5} dt \quad \text{where } a = \text{resultant linear head acceleration} \tag{1}$$

Impact location data recorded in vivo were grouped by azimuth and elevation into impact location bins (Fig. 4).

2.2 Laboratory Data Collection Methods. Laboratory data were collected according to current ($n=30$ impacts) and proposed ($n=54$ impacts) NOCSAE standard test protocols for football helmets. Three instrumented Riddell Revolution medium football helmets were tested in accordance with the current NOCSAE standards for football [2,28]. This testing protocol consisted of two 1.52 m drops for each helmet at front, side, front boss, rear boss, and rear impact locations. Helmets were only tested under the ambient temperature condition. The top impact location was not tested due to the lack of a comparable test location in the proposed standard. Facemasks were not affixed to the helmets for the linear drop test as per NOCSAE protocol. For each drop, acceleration data were sampled at 10,000 Hz and filtered using a CFC 180 filter to eliminate high frequency noise (as in Ref. [29]). Resultant CG linear acceleration was computed from the tri-axial acceleration measurements at the headform CG. Resultant acceleration was truncated to match the 40 ms window for HIT system in vivo data and GSI (Eq. (1)) was computed from the truncated acceleration signal.

The instrumented NOCSAE headform [2] was then removed from the NOCSAE drop test fixture and affixed to a standard 50th percentile male Hybrid III neck, which was mounted on a LI (Biokinetics and Associates Ltd., Ottawa, Canada) (Fig. 1). The same three helmets were equipped with Riddell G2EG-R facemasks. These facemask types were the most commonly selected by the players. The helmets were then placed on the headform and impacted at the front faceguard, front boss, side, rear boss, and rear impact locations at impactor velocities of 3 m/s, 6 m/s, and 9 m/s. The impact locations were selected in accordance with the proposed NOCSAE standard [15]. Additionally, all helmet samples were impacted at 11 m/s on the side, rear, and front faceguard locations. These locations were selected for the 11 m/s impacts because they were thought to be the least likely to cause permanent damage to the headform during high energy impacts. Impactor velocity was recorded by a light gate just prior to contact with the helmet. Data for each impact were collected at 10,000 Hz and filtered using a CFC 180 filter. Resultant CG acceleration was computed from the tri-axial acceleration measurements at the headform CG. Resultant acceleration was truncated to match the 40 ms window for HIT System in vivo data and GSI (Eq. (1)) was computed from the truncated acceleration signal.

2.3 Impact Location. The first goal of this study was to evaluate whether the impact locations prescribed in the proposed NOCSAE headgear standard were more representative of the location of impacts sustained during play than the impact locations prescribed in the current standard. Two metrics were used, as follows: (1) the distribution of all in vivo impacts by impact location in various GSI ranges, and (2) the percentage of impacts (as a function of GSI) that occurred in locations on the helmet, which are not testable according to the current and proposed standards.

In vivo head impacts were grouped by impact elevation, into seven equally spaced levels, −Q3 (the base of the head) to Q4 (the crown) as seen from a side view [30], and by impact azimuth, into eight equally spaced quadrants (front, left/right front boss, left/right side, left/right rear boss, and rear) as seen from a top view. Left and right side bins were grouped together, resulting in a total of 35 impact location bins (Fig. 4). The 35 impact location bins were classified as "tested" (location bin contains a required NOCSAE test location), "testable" (location bin does not contain a required test location but is within the testable limits for the random test location), and "nontestable" (location bin is not within the testable limits). Each location bin was classified based on both the current and proposed standards, e.g., a particular impact location bin may be testable according the current standard and nontestable according to the proposed standard. Data were also grouped by GSI into four bins: [0 200], [201 400], [401 600], and [>600]. To visually represent the data, heat maps, which projected the concentration of impacts onto a three-dimensional headform, were generated for each GSI bin (MATLAB, The MathWorks Inc., Natick, MA). Pearson's chi-squared (χ^2) was used to test the hypothesis that the frequency of in vivo impacts occurring in nontestable location bins was independent of the NOCSAE test protocol used to define the testable region (i.e., proposed or current NOCSAE standard). A χ^2 test for trends in binomial proportions [31] was used to test the hypothesis that the difference between the proportion of in vivo impacts in nontestable locations according to the proposed standard and the proportion in vivo impacts in nontestable locations according current standard was independent of the GSI bin.

2.4 Impactor Velocity. The second goal was to determine what impactor velocities for the proposed LI test method would recreate the 95th, 99th, and 99.9th percentile GSI values recorded in vivo, and whether these impactor velocities were independent of impact location. LI data were grouped into five subsets representing the five impact locations specified in the proposed standard (front, front boss, side, rear boss, and rear) and a least-mean-square exponential regression was fit to impactor velocity and GSI by Eq. (2)

$$\mathrm{GSI} = \beta e^{\alpha V} \tag{2}$$

where α and β are the regression parameters and V is the impactor velocity.

The exponential acceleration component of GSI (Eq. (1)) explains the exponential correlation between impactor velocity and GSI. For each impact location, the exponential regression was interpolated to determine the impactor velocities that would generate GSI values in the laboratory equal to the 95th, 99th, and 99.9th percentile GSI values for in vivo impacts to similar locations. Impacts to the Q4 elevation region (top of the head) were excluded from this analysis because the LI test apparatus cannot be configured for a direct impact to the top of the head (i.e., the regression for side LI data was interpolated at the 95th, 99th, and 99.9th percentile GSI values of in vivo data to the side region with the exception of data in Q4). Exponential regressions were linearized (impactor velocity versus log GSI) and modified analysis of covariance, which is a technique designed for analyzing regression lines of unequal slopes, was used to test for differences among the five regressions and to identify the domains over which differences in the regressions were significant according to the $\alpha = 0.05$ significance criterion [32].

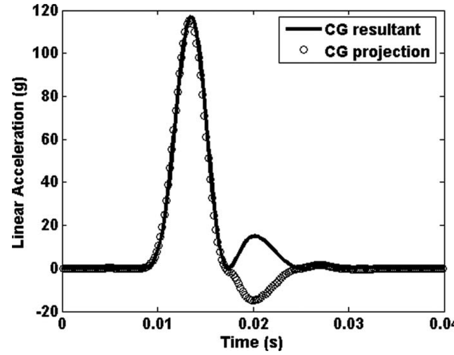

Fig. 5 CG resultant and CG projection for an in vivo rear location football impact

2.5 Impulse Response. Fifty-one impacts across three impact methods (LI, drop, and in vivo) were selected for lumped parameters systems modeling. Laboratory impact data for side, rear boss, and rear impact locations were grouped into two bins: pneumatic linear impactor (LI) or drop test (drop). These impact locations were selected because they are prescribed in both the current and proposed NOCSAE standards. Only LI data for the 9 m/s (8.92 ± 0.09 m/s, mean ± SD) impactor velocity condition was included for this analysis. This velocity was selected in order to generate GSI values in the range of the GSI values resulting from the 1.52 m linear drop tests (as prescribed in the current NOCSAE standard). In addition, a subset of 24 on-field impacts divided evenly among the three impact locations (side, rear boss, and rear) were selected for analysis. Impacts whose GSI values were closest to the mean laboratory GSI value for the particular impact location were selected. In order to most closely match the laboratory data, only in vivo impacts in elevation bins −Q1, Q1, or Q2 were considered.

For each impact a CG projection time series was defined as the projection of the CG resultant linear acceleration onto the impact direction vector, which is defined as the direction of maximum acceleration. The CG resultant or the magnitude of the CG linear head acceleration was always positive. The CG projection was either positive or negative and reflected both the magnitude and the direction of acceleration (Fig. 5).

MATLAB systems identification toolbox was used to create second- and third-order underdamped, continuous-time process models for each impact. These models were used to characterize each impact based on frequency domain parameters: the damping ratio (ξ), the natural period of oscillation (T_w), and, for the third-order model only, the time constants corresponding to the zero (T_z) and the third pole (T_p). The model input was an impulse of arbitrary magnitude at time zero, the time at which the absolute value of the CG projection exceeded 5g. The model output was the CG projection time series. Schematically, the models represented mass-spring-damper systems (Fig. 6) with parameters mass (M), damping factor (B) and spring constants (K, K_1 and K_2). The models were governed by the following equations of motion:

Equation of motion for second-order model:

$$Kx + B\dot{x} + M\ddot{x} = F \tag{3}$$

Equations of motion for third-order model:

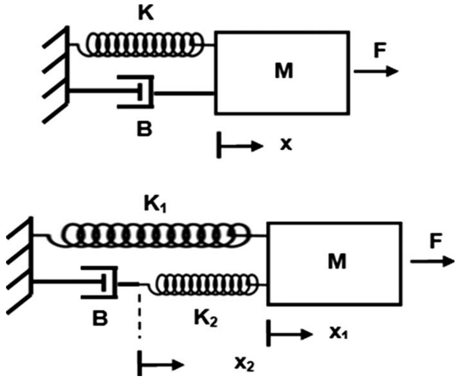

Fig. 6 Schematics of second-order (top) and third-order (bottom) mass-spring-damper models. The head/neck unit (*M*) is connected to a rigid body by springs (K, K_1, K_2) and a damper (*B*).

$$(x_1 - x_2)K_2 + K_1 x_1 + M\ddot{x}_1 = F \tag{4a}$$

$$(x_1 - x_2)K_2 - B\dot{x}_2 = 0 \tag{4b}$$

where M is the mass of the head/neck unit, K, K_1, K_2 are the spring constants, B is the damping factor, and F is the magnitude of the applied force impulse

From these governing equations of motion, generalized Laplace transfer functions were derived for the second-order (Eq. (5)) and third-order (Eq. (6)) models

$$G(s) = \frac{1}{1 + (B/K)s + (M/K)s^2} \tag{5}$$

$$G(s) = \frac{(K_2 + B \times s)}{(K_1 + K_2) + B(K_1 + K_2)s + (M \times K_2)s^2 + (M \times B)s^3} \tag{6}$$

The second- and third-order transfer functions were written in standard form (Eqs. (7) and (8), respectively) in order to utilize the MATLAB systems identification toolbox. For each impact, model parameters which maximized the fit, or percent of the CG projection that was explained by the model, were determined

$$G(s) = \frac{A}{1 + (2 \times \xi \times T_w \times s) + (T_w \times s)^2} \tag{7}$$

$$G(s) = \frac{A(1 + T_z \times s)}{(1 + (2 \times \xi \times T_w \times s) + (T_w \times s)^2)(1 + T_p \times s)} \tag{8}$$

where ξ is the damping ratio, T_w is the natural period of oscillation, T_p is the time constant corresponding to the third pole, and T_z is the time constant corresponding the zero.

Multivariate analysis of variance was used to identify mean differences in the model parameters (ξ, T_w, T_p, T_z) across categorical groups: impact location and impact method (LI, drop, in vivo). Parameter A, which is the static gain, was not relevant in this model because input force was arbitrarily assigned. Categorical groupings that showed significance according to the $\alpha=0.05$ significance criterion were explored further using one-way analysis of variance in order to determine which group differences were significant.

MATLAB was used for all data processing and statistical analysis. The statistical significance criterion was set at $\alpha=0.05$ a priori.

3 Results

From the in vivo data ($n=20{,}733$ impacts), the front impact bin represented the highest percentage (42.6%) of all impacts (Table 1). 54.7% of all impacts occurred to locations on the helmet, which are not testable according to the current standard and 37.7% of all impacts occurred to locations on the helmet, which are not testable according to the proposed standard. There were no significant changes in the difference between the percentage of in vivo impacts in nontestable locations as defined by the current and proposed methodologies as a function of GSI ($p=0.64$).

Table 1 Distribution of in vivo impacts by azimuth and elevation

	Azimuth location bin				
	Front	Front boss	Side	Rear boss	Rear
n	8833	3960	3380	2508	2052
%	42.6	19.1	16.3	12.1	9.9

	Elevation location bin						
	−Q3	−Q2	−Q1	Q1	Q2	Q3	Q4
n	6818	1882	1738	1685	1964	2649	3997
%	32.9	9.1	8.4	8.1	9.5	12.8	19.3

A total of 20,450, 193, 50, and 40 impacts were recorded in each of the four GSI bins, respectively. At all GSI levels, a lower percentage of in vivo impacts occurred to nontestable impact locations in the proposed standard than in the current standard ($p < 0.001$). For the lowest GSI group (0–200 GSI), the highest concentration of impacts occurred in the front of the head and for the highest GSI group (600+GSI) the highest concentration of impacts occurred in the top of the head (Fig. 7).

From the data collected in the laboratory using the NOCSAE LI apparatus, there was higher correlation between GSI and LI velocity for side, rear boss, and rear impact locations ($r^2 > 0.93$) compared with front and front boss locations ($r^2 < 0.83$) (Table 2). All five exponential regressions of LI velocity versus GSI are shown in Fig. 8.

The 95th, 99th, and 99.9th percentile GSI values for in vivo impacts to the rear azimuth region were 84.1, 299.1, and 1143.3, respectively. These values corresponded to LI velocities of 5.2 m/s, 7.8 m/s, and 10.5 m/s, respectively (Fig. 9). At 4 of 5 impact locations, the 99.9th percentile GSI value corresponded to an im-

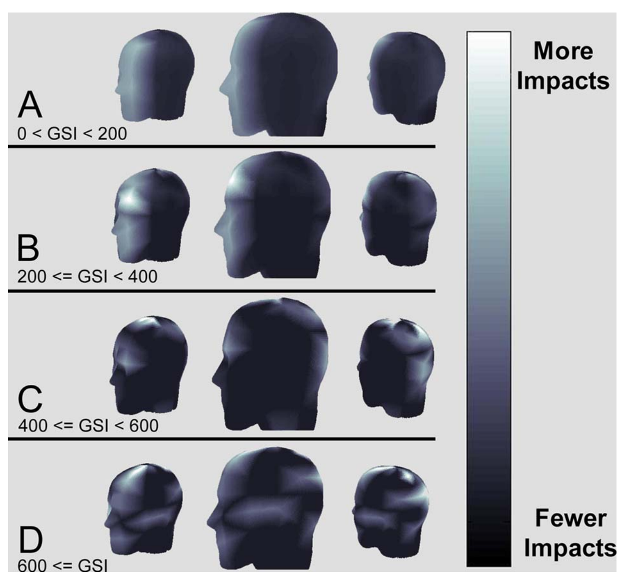

Fig. 7 Heat maps showing the concentration of impacts on the head surface: panel (*a*) contains data in the 0–200 GSI bin, panel (*b*) contains data in the 201–400 GSI bin, panel (*c*) contains data in the 401–600 GSI bin, and panel (*d*) contains GSI greater than 600

Table 2 Exponential regression coefficients and r^2 for impactor velocity versus GSI

	Front	Front boss	Side	Rear boss	Rear
α	0.39	0.67	0.49	0.50	0.46
β	5.51	0.91	6.84	5.65	7.31
r^2	0.77	0.82	0.99	0.94	0.95

pactor velocity greater than 9.5 m/s. The highest impactor velocity required to simulate the 99.9th percentile in vivo GSI value was 12.3 m/s for front (facemask) impacts (Table 3). Differences between regression lines for each impact location fitting log GSI to

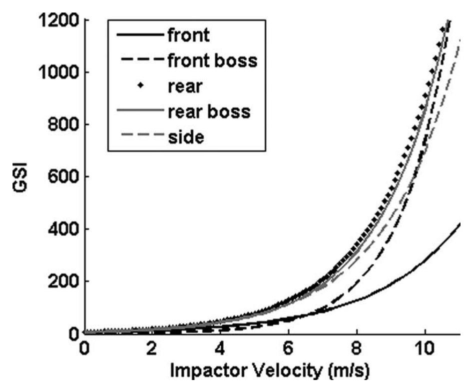

Fig. 8 Best fit exponential regressions of LI velocity versus GSI for all five impact locations. r^2 values and regression parameters are given in Table 2.

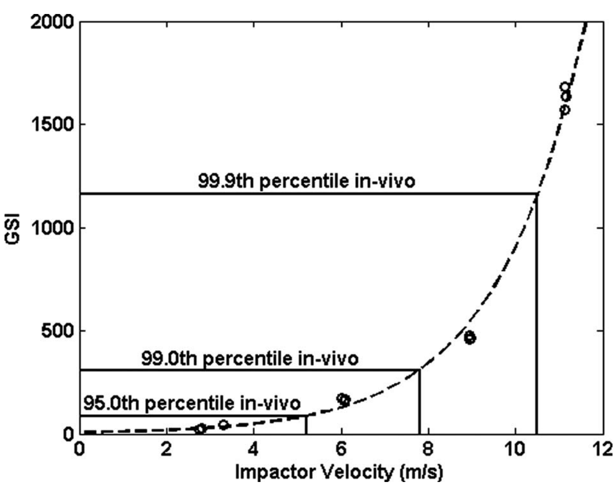

Fig. 9 LI velocity versus GSI for rear location LI impacts (shown as circles). The best fit exponential regression for this location was GSI=7.31$e^{0.46}$ (r^2=0.95, dashed line). Solid lines highlight the 95th, 99th, and 99.9th percentile of in vivo impacts to the rear azimuth region, excluding impacts to the Q4 elevation region (top of the head).

Table 4 Modified analysis of covariance significance domains

	Football				
	Front	Front boss	Side	Rear boss	Rear
Front					
Front boss	$V>9.0$ m/s				
Side	$V>3.8$ m/s	$V<7.9$ m/s			
Rear boss	$V>4.5$ m/s	$V<7.8$ m/s	None		
Rear	$V>3.6$ m/s	$V<8.6$ m/s	$V>7.6$ m/s	None	

"None" indicates that there were no significant differences between the two indicated regression lines over the domain 0–12 m/s.

impactor velocity were significant (Table 4). Regression lines for front and front boss impact locations were significantly different for impactor velocities >9.0 m/s, while rear boss and front boss regressions were only significantly different for lower energy impacts where impactor velocity <7.8 m/s. Differences between side and rear boss impacts, for example, were not significantly different over any domain within 0–12 m/s.

Across all impact types (in vivo, linear drop test, and LI), percent fit for the third-order process model was 78.0%, while the percent fit for the second-order model was 70.5% (Table 5). For in vivo impacts alone the percent fit for the second-order model was 59.7% compared with 74.8% for the third- order model. Based on these measures of fit the third-order model was selected for further analysis. Differences in the means of the model parameters (T_w, ξ, T_p, T_z) were significant by impact method $(p<0.001)$, but were not significant by impact location $(p=0.48)$. Differences in the mean values of the model parameters grouped by impact type were significant across all impact type combinations except for the differences in ξ for LI and drop data, differences in T_p for LI and drop data, and differences in T_p for LI and in vivo data, which were not significant at the $p<0.05$ significance level (means, standard deviations (SDs), p-values shown in Fig. 10).

4 Discussion

The objective of this study was to evaluate the recently proposed NOCSAE standard LI test method using in vivo measures of head impact acceleration recorded during American football play. The LI test method, proposed in 2006, was designed to more closely emulate on-field impacts believed to be responsible for mTBI [15]. There were two primary differences between the current drop test standard and the proposed LI standard. First, the proposed standard prescribes that a moving impactor strike a stationary headform attached to a slide table via a nonrigid neck while the current NOCSAE drop test utilizes a moving headform and rigid neck that strikes a rigid surface. Second, the proposed methodology requires that a facemask be affixed for all test impacts, including an impact directly to the center of the facemask, unlike the current methodology, which requires helmets to be tested without facemasks.

This study identified differences in impact locations for in vivo head impacts grouped by impact severity. The highest magnitude impacts tended to occur to the top of the head. These trends have been previously discussed by the authors [24]. In vivo impacts

Table 3 Interpolation of LI regressions at 95th, 99th, and 99.9th percentile GSI

	Front		Front boss		Side		Rear boss		Rear	
Percentile	GSI	Velocity (m/s)	GSI	Velocity (m/s)	GSI	Velocity (m/s)	GSI	Velocity (m/s)	GSI	Velocity (m/s)
95th	80.7	6.8	45.1	5.8	53.8	4.4	43.0	4.0	84.1	5.1
99th	212.1	9.3	120.5	7.3	144.4	6.5	151.4	6.5	299.1	7.7
99.9th	682.1	12.3	318.8	8.7	571.3	9.6	792.5	9.8	1143.3	10.5

Table 5 Systems model fit by impact type

Model	Model fit (mean ± SD) by impact type In vivo (% fit)	LI (% fit)	Drop (% fit)	All impacts (% fit)
Second-order	59.7 ± 19.5	77.0 ± 1.5	75.1 ± 3.5	70.5 ± 13.5
Third-order	74.8 ± 15.1	77.4 ± 4.3	80.7 ± 1.6	78.0 ± 9.1

occurred 37% more frequently on the regions prescribed in the proposed standard than those prescribed currently. This was largely accounted for by the inclusion of impacts to the facemask for the proposed standard. Contact area for any given impact may have spanned the border between two impact location bins. However, impact location was defined as a single point on the helmet reflecting the focal point of the impact, allowing each impact to be classified into a single impact location bin [18,23,24,30].

Exponential regressions of impactor velocity versus GSI demonstrated that if the proposed NOCSAE standard were implemented, impactor velocities as high as 10.5 m/s would be needed to simulate the highest energy impacts sustained in vivo. Specifically, the 99.9th percentile in vivo GSI value for rear location impacts was simulated in the laboratory by a 10.5 m/s LI velocity. The same value for front location impacts was 12.3 m/s but this value may not be reliable given the potential for penetration of the impactor into the facemask, which is not a phenomenon likely to be repeated on the field. We observed, using high speed video, several cases where the impactor penetrated the openings of the facemask in such a way as to unseat the helmet from the headform. This may be the cause of the lower r^2 value for the regression between LI velocity and GSI for facemask impacts. Modifications to the proposed test method would likely be required to eliminate the unrealistic condition of the impactor penetrating the facemask during testing.

For this study we selected to use GSI to quantify impact severity. The head injury criterion [33,34] is another commonly utilized head impact severity measure based on an integral of head acceleration versus time. Prior work [24] has demonstrated that for helmeted head impacts in American football the correlation between GSI and HIC is high ($r^2 > 0.9$). GSI was selected for analysis here because it is currently utilized as the pass/fail criterion in NOCSAE standards.

The impactor velocities required to simulate the highest GSI values recorded in vivo are consistent with high speed video reconstructions of impacts in the National Football League (NFL), which demonstrated closing speeds as high as 11.7 m/s for tackles in the open-field [35] and are further supported by the fact that a speed of 7.6 m/s is easily attainable even in full football gear (4.8 s, 40 yard dash). For comparison, a 1.52 m drop test (as in the current NOCSAE standard) results in a 5.4 m/s impact velocity.

There are several challenges associated with simulating high closing speed impacts using a drop test apparatus (as in the current NOCSAE standard). A drop height of over 6 m would be required to attain an 11 m/s impact velocity. Ceiling height limitations aside, a 6 m drop is not practical because of the energy transfer mechanisms for the drop test. During a linear drop, the kinetic energy in the system goes to zero (when the headform bottoms out and begins to rebound) while for the LI test some kinetic energy is maintained in the impactor and some is transferred to the sliding head/neck unit. As such, substantially higher impact energy is required to create a given headform GSI value for the proposed LI test method then for the current drop test method. For example, a 9 m/s impact with a 13.3 kg impactor (as prescribed in the proposed standard) corresponds to impact energy of 539 J and creates similar headform GSI values as a 1.52 m linear drop, which corresponds to 60 J, 71 J, and 88 J for small, medium, and large headforms, respectively.

The LI test apparatus was developed specifically to simulate high closing speeds, which are difficult to simulate using a linear drop test method, and thus the LI may be more representative of some in vivo head impacts. However, use of these high velocity impacts raises questions regarding the durability of the test apparatus, specifically, the NOCSAE headform and the interface between the Hybrid III neck and NOCSAE headform. Anecdotally, it is possible to shear a test headform off a Hybrid III neck during an 11 m/s impact.

Process models of second- and third-order were used to characterize in vivo and laboratory impacts based on frequency domain parameters including the damping ratio (ξ) and the natural period of oscillation (T_w). These parameters can be related to rheologic properties of the head/neck unit both in vivo and in the laboratory. In vivo, ξ can be related to the damping properties of the neck and shoulder musculature and to the helmet materials while T_w can be related to properties of neck stiffness and head mass as well as to an athlete's preparation for impact (i.e., braced for impacts with muscles tensed or blindsided/unprepared). In the laboratory, ξ can be related to the material properties of the Hybrid III neck for LI impacts or to the damping properties of the ground surface for drop impacts as well as to the helmet material properties. T_w can be related to headform mass and to the axial pretension or neck stiffness in the Hybrid III neck (for LI impacts)

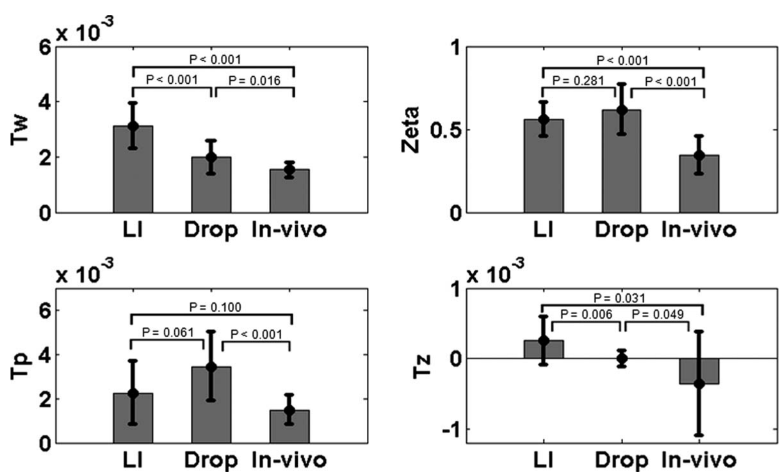

Fig. 10 Mean values for each model parameter (T_w, ξ, T_p, T_z) grouped by impact type. Errorbars are ±1 SD. *P*-values for each one-way analysis of variance are shown.

or to the stiffness of the headform mounting fixture (for drop impacts). Interestingly, the second- and third-order models both captured the variation in the headform CG projections for LI and drop impacts, while for in vivo impacts the third-order model accounted for noticeably more of the variation in CG projections. This may be explained by the introduction of the third pole (T_p), which allows the model to capture interactions between stiffness and damping parameters. While such interactions are expected for neck and shoulder musculature in vivo, they are not expected for the stiffness and damping properties of the Hybrid III neck.

The process models of third-order demonstrated that time series headform linear accelerations for the current standard, not the proposed standard, were most similar to in vivo accelerations. Specifically, the period of natural oscillation was approximately 2 times greater, and the damping factor was approximately 1.6 times greater, for LI impacts than for in vivo impacts. This suggests that increasing the stiffness of the Hybrid III neck would make the LI headform accelerations more similar to in vivo head accelerations. There are several alternative methods of decreasing the period of natural oscillation. The mass of the head/neck unit could be reduced but this would have to be coupled with a decrease in the damping coefficient (B) (Fig. 6) in order to ensure a net reduction in the damping factor (ξ), or the impactor surface could be made to be more compliant. However, the NOCSAE headform has been tuned to mimic the human head and the impactor surface has been tuned to simulate helmet-to-helmet impacts occurring in vivo. The parameters that have not been tuned for sport-related head impacts are the stiffness and damping properties of the neck. The Hybrid III anthropomorphic dummy was developed by the automotive industry for use in crash testing and was not designed to simulate human head/neck response for sport-related impacts. As such, the stiffness properties of the Hybrid III neck may not be representative of neck stiffness in vivo, particularly not when the athlete is braced for impact. In this case, the effective mass of the impacted head is increased by the tensed neck and shoulder musculature. Interestingly, the earlier model Hybrid II neck was considered to be too stiff for vehicle testing [36] but might be appropriate for helmet testing. The test methodology presented here could be used to evaluate changes in the LI test apparatus and to optimize these changes in order to create headform accelerations, which are most similar to in vivo head accelerations.

A benefit of the proposed NOCSAE LI test method compared with the current drop test method is that it generates linear and rotational headform accelerations. Currently, there are thought to be three primary mechanisms of brain injury: (1) focal type injuries resulting from direct impact and cranial fracture, (2) coup (the impacted side of the skull)/contrecoup (opposite side of the skull) injuries resulting from linear brain translation within the skull, and (3) diffuse axonal injury resulting from head rotational acceleration [37]. However, research has indicated that in helmeted sports, specifically football, head linear and head rotational accelerations are highly correlated [35,38]. Furthermore, helmets do not reduce head rotational accelerations [38], yet they are known to reduce the frequency of injury [1]. Accordingly, King et al. [38] conclude that brain injury is not caused by pure linear or pure rotational acceleration; rather it is caused by brain response to the complex interaction between linear and rotational head accelerations. Until a feasible and cost effective way to mitigate rotational accelerations via improvements in helmet design has been demonstrated and the relative importance of protection from linear versus rotational acceleration has been determined, a test methodology that evaluates helmets based on headform rotational acceleration may not provide additional information for the purposes of differentiating helmets with respect to mTBI prevention.

Future work should include studies on rotational accelerations and associated concussive injuries in vivo, in an effort to understand the relative importance of linear and rotational head acceleration mitigations. However, it is widely accepted that both linear and rotational accelerations contribute, at some level, to concussive injury. As such, efforts to develop cost effective means of mitigating rotational accelerations via helmetry should be a high priority even before the relative risk of rotational acceleration in humans has been precisely quantified.

There are several limitations of this study. Multiple facemask types were worn during play but only one facemask was selected for laboratory testing; 53% of players wore the selected facemask (or a similar dual cross bar, carbon steel facemask). This limitation likely only affected the exponential regression between impactor velocity and GSI for front impacts. This study was limited to one helmet model. Consistent helmet selection was beneficial when evaluating head/headform response across the various test methods; however, the study could be repeated for alternate helmet types such as Riddell (Chicago, IL) VSR4 football helmets. In fact, the proposed LI test method was not designed specifically for football. A similar study could be conducted using instrumented ice hockey helmets. This study was limited to collegiate male athletes. Head/neck properties vary by gender and age and thus it is possible that head accelerations would be dissimilar for youth athletes and/or female athletes. Specifically, impulse response characteristics of in vivo head impacts and impactor velocities that represent the 99.9th percentile in vivo GSI values may vary with gender, age, and level of play. Finally, the damping factor (ξ) and period of natural oscillation (T_w) were not determined for impacts across the full range of GSI values sustained in vivo. Impacts were selected for analysis such that differences in GSI between the test methodologies would be minimized. Further analysis should characterize impacts of different GSI levels by the same model parameters (ξ and T_w).While this methodology is helpful in designing a helmet test apparatus and determining appropriate impact energy for a standards test, it does not address the issue of selecting an appropriate impact tolerance threshold for mTBI and a pass/fail criterion for the proposed standard.

The knowledge gained from this study may provide the basis for developing a humanlike neck designed specifically to mimic direct head impacts in sports. In developing this neck it may be beneficial to provide a means to vary and to monitor the neck stiffness. Monitoring the neck stiffness would ensure laboratory-to-laboratory repeatability and varying the neck stiffness may facilitate more appropriate test conditions for youth helmets. Further testing of the LI test apparatus with head/neck modifications, using the methodology presented herein, is recommended.

Conflict of Interest: Authors R. G., J. C., and J. C. have a financial interest in HIT System technology used to record in-vivo head impacts for this study.

Acknowledgment

This work was funded in part by the National Operating Committee for Standards in Athletic Equipment (NOCSAE) and the National Institutes of Health (NIH) (Grant No. R01HD048638). The HIT System was developed by Simbex (Lebanon, NH) through a NIH Small Business Innovation Research (SBIR) award (Grant No. R44HD40473). In addition, this work was supported by research and development funding from Riddell, Inc. (Chicago, IL). Analog Devices (Cambridge, MA) provided sensors and expertise in sensor integration. We appreciate and acknowledge our colleagues Tom McAllister, MD and Tina Duhaime, MD from Dartmouth Hitchcock Medical Center, Lebanon NH, Mel Cook, Southern Impact Research Center, Rockford TN, for performing laboratory helmet testing, Jeff Frechette and Scott Roy, Dartmouth College Sports Medicine, for invaluable insight and daily monitoring of the in vivo data acquisition system, and the engineering team at Simbex, including Jonathan Beckwith, Siena Calabro, Lindley Brainard, and Dan Leonard, for manufacturing and supporting the instrumented helmets. Additionally, we would like to thank the players, athletic training staff, equipment staff, and coaches from the Dartmouth College football team for their ongoing support and cooperation.

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