Title: The influence of head impact threshold for reporting data in contact and collision sports: System review and original data analysis.

Running title: Reporting impact data in sport.

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Running title: Reporting head impact data in sport.

Abstract

Background: Head impacts and resulting head accelerations cause concussive injuries. There is no standard for reporting head impact data in sports to enable comparison between studies.

Objective: To outline methods for reporting head impact acceleration data in sport and the effect of the acceleration thresholds on the number of impacts reported.

Methods: A systematic review of accelerometer systems utilised to report head impact data in sport. Calculation of the effect of using different thresholds on a set of impact data from 38 amateur senior rugby players in New Zealand (NZ) over a competition season.

Results: Of 52 studies identified, 42% reported impacts using >10*g* threshold. Studies reported descriptive statistics as mean ±standard deviation, median, 25th to 75th interquartile range, and 95th percentile. Application of the varied impact thresholds to the NZ data set resulted in 20,687 impacts >10*g*; 11,459 (45% less) impacts >15*g*; and 4,024 (81% less) impacts >30g.

Discussion: Linear and angular raw data were most frequently reported. Metrics combining raw data may be more useful, however validity of the metrics has not been adequately addressed for sport. Differing data collection methods and descriptive statistics for reporting head impacts in sports limits inter-study comparisons. Consensus on data analysis methods for sports impact assessment is needed, including thresholds. Based on the available data, the 10*g* threshold is the most commonly reported impact threshold and should be reported as the median with 25th and 75th interquartile ranges as the data is non-normal distributed. Validation studies are required to determine the best threshold and metrics for impact acceleration data collection in sport.

Conclusion: Until in-field validation studies are completed, it is recommended that head impact data should be reported as median and interquartile ranges using the 10*g* impact threshold.

1. Introduction

1.1 Head impacts cause injury – evidence

Known as the 'silent injury',[1] and often reported by the media and sporting circles as a 'knock to the head',[2] sport-related concussions (hereafter called 'concussion') are a subset of mild traumatic brain injuries (mTBIs)[3] and have become an increasingly serious concern for all sporting activities worldwide.[4-6] Research into concussions[7] has increased over the years leading to greater insight into the causes and the effects of these injuries. Research[8-27] has sought to better determine the head linear and rotational accelerations involved in concussion injuries through the use of telemetry. By adapting radio-telemetry that was utilised for astronauts,[28] the telemetry system has been in use since 1961 for the recording of impacts for football players and concussions[29] that have occurred.

1.2 A cumulative head impact threshold may be related to concussion

The immediate and long term effects of multiple and repeated blows to the head that athletes receive in contact sporting environments are a growing concern in clinical practice.[30, 31] Concern has grown about the effects of subconcussive impacts to the head and how these impacts may adversely affect cerebral functions.[30-32] Subconcussive events are impacts that occur where there is an apparent brain insult with insufficient force to result in the hallmark signs and symptoms of a concussion.[31, 33, 34] Although subconcussive events do not result in observable signs and apparent behavioural alterations,[35, 36] they can cause damage to the central nervous system and have the potential to transfer a high degree of linear and rotational acceleration forces to the brain.[37] Proposed decades previously,[38, 39] exposure to repetitive subconcussive blows to the head may result in similar, if not greater damage than a single concussive event[33] and may have cumulative effects.[40]

Participants can be exposed to a high number of impacts per season.[32] It has been suggested[41, 42] that brain injuries come from concussive events and also from the accumulation of subconcussive impacts that result in pathophysiological changes in the brain. As subconcussive impacts do not result in observable concussion related signs and symptoms, these are often not medically diagnosed. The accumulation of subconcussive blows can result in neuropsychological changes.[30, 31, 42-46] However, similar to the literature focused on concussion and mild traumatic brain injury (mTBI), the literature on subconcussive head trauma is limited.[47] What is not known is the number of head impacts and their intensity that might lead to concussion (i.e. a concussion cumulative threshold). The injury threshold is likely to be different for each person given the multifactorial nature of injuries, as per other thresholds for injuries to tendons, ligaments, muscle and bone. If a threshold could be determined, then players could be monitored to reduce their potential risk for concussion injury – akin to cricket monitoring players loading to the body during bowling events via the number of overs in an attempt to reduce the risk of back stress fractures.[48]

1.3 Impacts can be measured with a number of technologies

Head impact dynamics have been analysed through the use of video analysis,[8] in game measurements,[20-25, 27, 49-52] numerical methods[9-12] and reconstructions using anthropometric test devices[13-19] in helmeted sports such as American football[20-23] and ice hockey[24, 25] and in un-helmeted sports such as soccer[26] and rugby union.[27]

The on-field assessment of head impacts has been captured with a head impact telemetry system (HITS) (Simbex, LLC, Lebanon, NH) using helmet mounted accelerometers enabling determination of the head linear and rotational accelerations in American football,[21, 23, 49, 53-55] ice hockey[24, 25] and in a headband in youth soccer.[26] The data collected through the HITS has enabled analytical risk functions,[16, 51, 56, 57] concussion risk curves,[51] and risk weighted exposure metrics[58] to be developed further assisting in the identification of sports participants at risk of concussive injuries. More recently, instrumented mouthguards known as XGuard (X2biosystems, Inc., Seattle, WA, USA) have documented head impacts in rugby union.[27]

1.4 Thresholds have differed for reporting impact data in contact and collision sports

Although there is an increasing amount of published literature reporting impact accelerations to the head in the sporting environment, there is less attention focussed on identifying what is a subconcussive impact and where this occurs. Studies[55, 59, 60] have been conducted reporting the impacts absorbed by the head during activities undertaken daily. Although impacts to the head and body under 10*g* have been reported[55], these activities such as walking, jumping, running and sitting are considered to be non-contact events.[21, 61] However, impacts greater than 10*g* occurring from contact events that do not result in acute signs or symptoms of concussion, are identified as subconcussive impacts.[43]

1.5 To enable comparison of studies, a consistent threshold for reporting is needed

Head impact data are essential to understand the biomechanics of head injury to develop potential injury prevention strategies. There is currently no standard for reporting head impact data to enable comparison between studies. Currently the use of accelerometers may not necessarily provide the meaningful inter-study comparisons that are sought due to data collection, processing and methodologies not being standardized.[62] Studies utilising different impact thresholds have proposed varying conclusions based on the methodological and reporting approaches undertaken.

1.6 Aim of the study

The rationale for this study is based on questions around the magnitude of a single impact that may result in concussion, the number of impacts needed to result in signs and symptoms of concussion, and individual player differences that might affect injury tolerance levels for concussion. Given head impacts are likely to cause concussive injury, and the number of head impacts may be related to a potential concussion threshold (i.e. a cumulative threshold), the number of head impacts should be monitored in players. However, given impacts can be measured with a number of technologies (e.g. instrumented behind the ear patches, mouthguards, head gear), and thresholds have differed for reporting impact data in contact and collision sports, a threshold for reporting impact data in sport is needed to enable comparison of studies.

Therefore the aims of this study were to: a) outline the methods for reporting head impact data in sport; and b) to identify the effects of the acceleration threshold on the impacts reported.

2. Methods

To outline methods for reporting head impact data, a systematic review of the literature was conducted. The guideline for reporting observational studies (MOOSE: Meta-analysis Of Observational Studies in Epidemiology)[63] was followed for the empirical literature evidence included in this study. The MOOSE checklist contains specifications and guidelines for the conduct and review of the studies. To evaluate the effects of acceleration thresholds on the number of impacts reported, variable thresholds were applied to head impact data obtained from 38 senior amateur rugby union players during 19 matches in New Zealand.[27]

2.1 Literature review to identify thresholds for reporting head impact data in contact and collision sport

2.1.1 Search strategy for identification of publications

A total of 53,183 studies available online from Jan 1990 to June 2015 identified through the SCOPUS (n=10,080), SportDiscus (n= 1,185), OVID (n= 9,724), Science Direct (n= 27,798) and Health Sciences (n= 4,376) databases were screened for eligibility (see Fig. 1). The keywords utilized for the search of relevant research studies included combinations of 'head impact telemetry system*', 'HITS', 'concussion', 'impact*', 'traumatic brain injury', 'chronic traumatic encephalopathy', 'angular', 'linear', 'rotational', 'acceleration', 'biomechanics', 'head acceleration' and 'risk'. An example of the Health Sciences search strategy is provided in the Electronic Supplementary Material (ESM) S1. Searches were limited to 'English language' and 'humans' only. The references of all relevant articles were searched for further articles. All publications identified were initially screened by publication title and abstract to identify eligibility. In cases of discrepancies of eligibility another author assessed the publication to screen for eligibility.

To establish some control over heterogeneity of the studies,[63] inclusion criteria were established. Any published study or book that did not meet the inclusion criteria was excluded from the study. Publications were included if they reported head impact biomechanics and met the following inclusion criteria:

(i) The study was published in a peer reviewed journal or book; and

(ii) The study reported the biomechanics of impacts to the head in a sporting environment; and

(iii) The study addressed one or more of the keywords relating to this study.

Reviewed studies were excluded from this review if it was identified that the publication:

(i) Was unavailable in English; or

(ii) Did not provide additional information specifically addressing areas relating to this study;

(iii) Was a case study; or

(iv) Reviewed head impact studies.

2.1.2 *Assessment of publication quality*

The 52 studies[10, 12, 16, 20-27, 32, 37, 42, 49, 51-54, 57, 58, 61, 64-92] meeting the inclusion criteria (see Table 1) were assessed for quality by two of the authors on the basis of the MOOSE[63] published checklist. Heterogeneity of the studies included in the literature review was expected as there might be differences in the study design, population and outcomes.[63] As a result of the MOOSE[63] checklist, the studies included had a median score of 4.8/6.0 with a range of 4.0-5.0.

2.2 Application of head impact thresholds identified from the literature to the rugby head impact data set

The data set, used for the application of the head impact thresholds identified from the literature review, was from 38 amateur rugby union players who wore instrumented mouthguards over a season of matches.[27] The raw data set was filtered by linear acceleration thresholds at increments of 1*g* to establish the percentage of impacts removed at each threshold from 10.0*g* to 30.0*g*. This percentage was then used to calculate the possible number of impacts removed for the impact thresholds used in the different studies reviewed.

All data estimations were calculated on an Excel spreadsheet. The data were analysed using SPSS v22.0.0 (SPSS Inc.) and, as the data were non-normally distributed (Shapiro-Wilk test $p<0.001$), data were analysed using a Friedman repeated measures ANOVA on ranks. Post hoc analysis with Wilcoxon signed-rank tests was conducted with a Bonferroni correction applied. Statistical significance was set at $p<0.05$. The estimated number of impacts were calculated by dividing the number of reported impacts by the estimated percentage of impacts removed at the different thresholds. The estimated total number of reported impacts were subtracted from the reported number of impacts to identify the possible number of impacts removed from the data set e.g. Number of impacts reported = 161,732;[75, 76] Impact threshold = 14.4*g*; Based on New Zealand rugby union dataset for 20,687 impacts recorded at 10.0*g* when reassessed at 14.4*g* there were 12,091 impacts. A total of 8,569 impacts were removed or 42% of the data set (see Fig 2). Therefore 161,732 (number of impacts reported) ÷ 42% (percentage of impacts removed at 14.4*g*) gave a possible total number of impacts at the 10*g* threshold of 385,076. The possible total number of impacts removed from the dataset was 223,344 (i.e. 385,076 – 161,732 impacts).

3. Findings

3.1 Literature review

A total of 52 publications were identified that reported head impacts and met the inclusion criteria. Studies reported impacts to the head via technology in American football,[21, 22, 26, 37, 49, 51, 52, 54, 58, 61, 64, 65, 67, 69, 73, 75, 76, 79, 80, 85] ice hockey,[24, 25, 71, 72, 84, 92] soccer,[26] rugby union[27] and mixed martial arts and boxing.[64]

3.1.1 Impact threshold

Studies utilised different data impact acceleration thresholds (see Table 1): 42% of studies[21, 22, 24, 26, 27, 49, 52, 54, 61, 64-74] used 10*g*; 18% of studies[20, 23, 42, 51, 53, 58, 75-78] used 14.4*g*;10% of studies[37, 79-83] used 15*g*; 4% of studies[25, 84] used 20*g*; 2% of studies[85] used 30*g*; 4% of studies[32, 86] reported impact data within 10*g* to 60*g* and greater than 90*g*. Four studies[10, 12, 16, 87] (8%) were reconstruction studies from video analysis but were included as they reported impact biomechanics. Six studies[57, 88-92] (12%) did not report the impact threshold but did report head impact biomechanics. One study[64] (2%) used a 7*g* and 10*g* threshold with different sporting activities.

3.2 Acceleration raw data and metrics

Apart from raw resultant linear accelerations[32, 49, 52, 61, 65, 68, 85, 86, 91] (reported in 91% of studies) and rotational acceleration data[10, 51] (reported in 76% of studies),[12, 16, 20-24, 26, 27, 37, 54, 57, 58, 66, 69-83, 87-90] several head impact derived variables were reported such as the Gadd Severity Index (GSI),[93] the Head Impact Criterion (HIC),[94] Head Impact Telemetry Severity Profile (HIT_{SP})[90] and the Risk Weighted Cumulative Exposure (RWE)[58] metrics.

Three (4%) of the studies[26, 49, 75] reported the Gadd Severity Index (GSI). In 1966, Gadd[93] proposed the GSI head injury severity index based on the Wayne State Tolerance Curve (WSTC). Developed from animal and cadaver impact data, the GSI simplified the WSTC by taking into consideration the shape of the linear acceleration time history, providing a weighting factor of 2.5 enabling the whole body acceleration data to be plotted on log-log coordinates along a straight line. The critical value of the GSI is 1,000. If the GSI is less than 1,000 then the head impact is considered probabilistically safe. The GSI is used to quantify severe skull fractures and brain injury risk but is not recommended for use to quantify a risk of concussion.[95] A concern of the GSI is that it can give unrealistically high values for impacts that have a much longer pulse duration.[96] The mathematical expression for the GSI is:

$$GSI = \int_0^T a(t)^{2.5} dt$$

where *a* is the 'effective' acceleration (thought to have been the average linear acceleration) of the head measured in terms of *g*, the acceleration of gravity, and *t* is the time in milliseconds from the start of the impact.[97]

In 1971 a modification of the Gadd Severity Index, the Head Injury Criterion (HIC), was proposed[94] to focus the severity index on that part of the impact that was likely to be relevant to the risk of injury to the brain. This was done by averaging the integration of the resultant acceleration/time curve over whatever time interval yielded the maximum value of HIC. Because this varies from one impact to another, the expression for the modified index simply refers to times *t*1 and *t*2. The HIC is computed based on the following expression:

$$HIC = \left[\frac{1}{t_2 - t_1}\int_{t_1}^{t_2} a(t)dt\right]^{5/2}(t_2 - t_1)$$

where t_2 and t_1 are any two arbitrary time points during the acceleration pulse. Acceleration is measured in multiples of the acceleration of gravity [g] and time is measured in seconds. The resultant acceleration is used for the calculation. The US National Highway Traffic Safety Administration (NHTSA) requires t_2 and t_1 not to be more than 36 ms apart (thus called HIC_{36}) and the maximum HIC_{36} not to exceed 1,000. In 1998[98] the NHTSA introduced the HIC_{15} where t_2 and t_1 was not to be more than 15 ms apart and the maximum HIC_{15} was not to exceed 700. In a numerical study[99] it was estimated that a mild Traumatic Brain Injury (mTBI) tolerance for the HIC_{15}, where there is a 25%, 50% and 75% likelihood of an mTBI occurring, had HIC_{15} values of 136, 235 and 333 respectively. Only two studies[24, 76] (4%) reported HIC_{36} with ten studies (18%) reporting the HIC_{15}.[10, 12, 16, 24, 26, 49, 68, 75, 76, 90]

In 2008,[90] the principal component score (PCS), a weighted sum of linear acceleration, rotational acceleration, HIC and GSI, with objectively defined weights, was published. Now more commonly termed the Head Impact Telemetry Severity Profile (HIT_{SP}), the HIT_{SP} is a weighted composite score including linear and rotational accelerations, impact duration, as well as impact location. The resulting formula is:

$$HIT_{SP} = 10x([0.4718\ x\ sGSI + 0.4742\ x\ sHIC + 0.4336\ x\ sLIN + 0.2164\ x\ sROT] + 2)$$

where sX = (X-mean[X])/(SD[X]), LIN = linear acceleration, ROT = rotational acceleration, HIC = head injury criterion, and GSI = Gadd Severity Index. The offset by 2 and scaling by 10 generates HIT_{SP} values greater than 0 and in the numerical range of the other classic measures studied. A HIT_{SP} score of 63 or greater is reported to be an indication there is a 75% risk of a concussive injury occurring.[90] More than a quarter (30%) of the studies[21, 37, 69, 71-74, 77, 81-83, 90] reported the HIT_{SP}.

In 2013, a novel cumulative exposure metric, the Risk Weighted Cumulative Exposure (RWE) equation was developed[58] with four previously published analytical risk functions. The four different analytical risk functions were the linear resultant acceleration,[16, 56] rotational resultant acceleration[51] and combined probability (linear and rotational) resultant accelerations.[57] These risk functions were utilised to elucidate individual player and team-based exposure to head impacts. The RWE equations comprise of a_L as the measured peak linear acceleration, a_R as the measured peak rotational acceleration, and n_{hits} as the number of head impacts in a season for a given player.

Risk function(s)	Equation
Linear[12, 13]	$RWE_{Linear} = \sum_{i=1}^{n_{hits}} \boldsymbol{R}(\boldsymbol{a}_L)i$
Rotational[51]	$RWE_{Rotational} = \sum_{i=1}^{n_{hits}} \boldsymbol{R}(\boldsymbol{a}_R)i$
Combined Probability[57]	$RWE_{CP} = \sum_{i=1}^{n_{hits}} \boldsymbol{CP}(\boldsymbol{a}_L, \boldsymbol{a}_R)i$

Logistic regression equations and regression coefficients of the injury risk functions utilised in the prediction of injury, where α and β are the regression coefficients and x is the measured acceleration for the linear and rotational risk functions.[58]

Logistic Regression equation	Risk Function	Regression coefficients
$R[a] = \frac{1}{1+e^{-\alpha+\beta x}}$	Linear[12, 13]	α = -9.805, β = 0.0510
	Rotational[51]	α = -12.531, β = 0.0020
$\mathrm{CP} = \frac{1}{1+e^{-(\beta_0+\beta_1 a+\beta_2 \alpha+\beta_3\, a\alpha}}$	Combined Probability (CP)[57]	β_0 = -10.2, β_1 = 0.0433, β_2 = 0.000873, β_3 = -9.2E-07

β_0, β_1, β_2 and β_3 are regression coefficients, a is the measured linear acceleration, and α is the measured rotational acceleration for the combined probability risk function. The three metrics provided as a result of these equations are for linear (RWE_{Linear}), rotational ($RWE_{Rotational}$) and combined (linear and rotational) probability (RWE_{CP}). Only one study[58] has reported the RWE

In an attempt to delineate injury causation and to establish a meaningful injury criterion through the use of actual field data, Zhang et al.[12] proposed tolerance levels for human head injury based on input kinematics scaled from animal data and non-injurious volunteer test results. Injury predictors and injury levels were analysed based on resulting brain tissue responses and these were correlated with the site and occurrence of a concussion occurring. The calculated sheer stress around the brainstem region could be an injury predictor and statistical analyses were performed to establish a brain injury tolerance level. As a result of the analyses undertaken, and based on linear logistic regression analyses, it was reported[12] that the maximum resultant translational acceleration at the center of gravity (CG) of the head was estimated to be 66*g*, 82*g* and 106*g* for a 25%, 50% and 80% probability of sustaining an mTBI respectively.

For resultant rotational acceleration at the CG of the head this was estimated to be 4,600 rad/s^2, 5,900 rad/s^2 and 7,900 rad/s^2 for a 25%, 50% and 80% probability of sustaining an mTBI respectively. The estimated HIC_{15} thresholds were 151, 240 and 369 for a 25%, 50% and 80% probability of sustaining an mTBI. These thresholds are considerably less than the HIC_{15} limit of 1,000 for sustaining a serious brain injury. If the head was exposed to a combined translational and rotational acceleration with an impact duration between 10 to 30 ms, the suggested tolerable reversible brain injury was 85*g* (translational acceleration), 6,000 rad/s^2 (rotational acceleration) and HIC_{15} value of 240. It was reported that these values may change as more human data become available but to date no published updates of these values have been available.

Although other variables have been proposed (Generalised Acceleration Model for Brain Injury Threshold (GAMBIT),[14, 64, 100] and Head Impact Power (HIP);[101] these were not utilised in any studies reporting head impacts in contact sport.

Nearly all of the studies reviewed identified the number of impacts that were recorded, however 4% studies reported impacts in matches only, 23% recorded impacts for both match and practice activities, and 55% combined both match and practice activity impacts. The remaining 15% of studies reviewed reported on impacts above 90*g* or were reconstruction of impacts from video analysis The number of impacts ranged from 480 impacts from 22 players in Pop Warner American football[85] to 486,594 impacts from 450 players in collegiate American football and ice hockey[89] (see Table 1).

Over half (52%) of the studies[10, 12, 16, 22, 23, 27, 37, 49, 58, 61, 65, 66, 69-75, 79, 82-86, 91, 92] reported the impact biomechanics data as mean ± standard deviation (±SD). Some studies[23, 25, 58, 64, 73, 75, 82] (22%) also reported the head impacts as median, but not all[23, 73] (4%) included the interquartile ranges (IQR) for the data. Of the studies that reported the impact biomechanics by the median, only 7% reported the IQR. Most of the studies reporting the median also reported the 95th percentile of the impacts. Other data reporting methodologies utilised within the data sets reviewed were the median of the 95th percentile,[21] the 98th,[82, 90] 99th,[82, 90] and 99.5th[82] percentiles. Fourteen percent of studies also included lower and upper limits[61, 71, 72, 74] for the range of impacts,[24, 89] and the mean range[85] of the impacts. Less than a quarter of studies (23%) reported their impacts as x, y, z axis data,[22] +1SD,[52] Cumulative Distribution Functions (CDF),[54, 58] percentage of impacts,[21, 53] and the impact duration (ms).[16, 75, 76, 80, 81] In addition to the impact biomechanics being presented by various methodologies, 14% of studies[12, 27, 37, 69, 74, 79, 91] also incorporated impact tolerances and impact severity levels.

3.3 Application of head impact thresholds to the rugby head impact data set

By utilising data from a previously published study[27] that used the 10*g* impact threshold, data were re-extracted at differing impact thresholds from 10*g* to 30*g*. By adjusting the impact threshold (see Fig. 2) the number of impacts decreased as the impact threshold increased (see Table 2). There were significant differences observed

(p<0.05) for each of the different acceleration thresholds for the number of impacts reported, the mean, median and the 95th percentile when compared with the impacts at the 10*g* linear acceleration threshold (see Table 2).

Based on the differences observed in this study, at the 14.4*g* threshold there could have been as many as 42% of the impacts recorded not being reported. As a result, studies[20, 23, 51, 58, 75-82, 85] using impact thresholds above 10*g* may have removed 2,100 to 206,573 impacts. At the 30*g* impact threshold it can be estimated that 80 to 85% of impacts were not reported.[85] Again, based on the differences observed in this study it is possible that each player in the Pop Warner study[85] may have experienced a cumulative total of 1,885 impacts above 10*g*. Although the impacts may not have been recorded, the players may well have been exposed to this number of impacts between 10*g* and 30*g*. The differences between impacts reported and the possible number of impacts (480 vs. 2,365) may result in an underestimation of the exposure risk to these players to subconcussive impacts.

4. Discussion

This study undertook to review the methods for reporting head impact data in sport and to outline the effect of various acceleration thresholds on the number of impacts reported. A consensus on a threshold for reporting data is important given the variation in conclusions that may be drawn if the same dataset is used with different thresholds, as identified by our application of the range of thresholds from prior literature applied to a New Zealand rugby union head impact data set. A standard threshold for head impact data is important given possible monitoring of player head impact acceleration data in the hope of identifying a cumulative threshold for concussion from subconcussive impacts.

The discussion surrounding subconcussive impacts has become popular.[32, 41, 43, 83, 102, 103] Initially the term subconcussive impact described an impact that did not result in severe, noticeable symptoms, especially loss of consciousness[102] However, recently, subconcussive is a term used to describe an asymptomatic non-concussive impact to the head.[32, 41, 43, 83, 103] The issue relating to the effects of subconcussive impacts is controversial as researchers and clinicians are divided on the true effects.[30-32, 42, 45, 104] Some research[32, 104] has reported that these impacts have minimal effect on cognitive functions, while others[30, 31, 42, 45, 46] have reported these impacts to be detrimental to cerebral and cognitive functions. To date, there is a paucity of evidence to identify the impact acceleration that is adequate to produce a non-structural brain injury associated with the neuronal changes of concussion.[30]

Animal models display metabolic changes associated with concussion, which may be similar in subconcussive impacts.[105] To research subconcussive impacts in isolation is challenging and there are, to date, no reports on animal models or other reliable methodologies that have been successful at identifying these impacts[105] Brain injury may occur from concussive events as well as from an accumulation of subconcussive impacts.[41] The effects of concussive events and multiple subconcussive impacts have been associated with long term progressive neuropathologies and cognitive deficits.[43, 106-108] Longitudinal impact monitoring at the level where these subconcussive events are beginning to occur is important, and a standard threshold needs to be established.

4.1 *What threshold should be used to monitor head impacts?*

Impacts <10*g* of linear acceleration have been considered negligible in regards to impact biomechanical features. The <10*g* impact threshold has been used in research to eliminate head accelerations from non-impact events such as jumping and running.[21, 55, 61] The inclusion of these non-impact events to head trauma make it difficult to distinguish between head impacts and voluntary head movement[109] and eliminating these will help identify the true extent of the number of impacts that do occur from sports participation. A suggestion for this may be to report the distribution of the impacts by the various resultant linear accelerations using a frequency analysis and reporting quartile ranges i.e. 25th and 75th interquartile range. This may assist in identifying where the most frequent resultant linear accelerations occur in the different sports. Consensus for the impact threshold will need to be established, and should be based on validation studies to determine the best impact threshold for various sports and injury outcomes. Biomechanical modelling of impact forces and brain movement would be needed to identify likely impact thresholds for injury, as well as in-field validation studies using prospective monitoring of players during tackles and impacts with the ground. As there is no established criterion for reporting head impact biomechanics, and the majority of studies (42%)[21, 22, 24, 26, 27, 49, 52, 54, 61, 64-74] reported the resultant linear acceleration threshold at 10*g*, then future studies should report all impacts above the 10*g* resultant linear acceleration threshold.

4.2 *What descriptive statistics should be used to report head impact biomechanics?*

There were a variety of descriptive statistics used in the reporting of head impact biomechanics in the reviewed studies which limits inter-study comparisons. Although more than half (52%) of the studies reviewed[10, 12, 16, 22, 23, 27, 37, 49, 58, 61, 64-66, 69-75, 79, 82-86, 91, 92] reported their results by means and standard deviations, the use of these statistics may not accurately represent the true centre of the data. By reporting the mean value of the data set, this method is subject to extreme values (i.e. outliers) such as those in skewed datasets. The use of the mean is only appropriate if the dataset is normally distributed. In non-normal distributed data, the median is the most useful for describing the center of the data. Of the studies[23, 25, 58, 73, 75, 82] reviewed (22%) that reported the results by the median would more accurately have identified the center of the dataset. The New Zealand senior amateur head impact data were non-normally distributed (i.e. not symmetrical) therefore the use of descriptive statistics that can account for this skewness needed to be considered. To enable inter-study comparisons, and until a consensus is established for the reporting of head impact biomechanics, future studies should report the median [25th and 75th interquartile ranges] for all head impact biometrics.

4.3 What acceleration metrics should be used to monitor head impacts?

It has been suggested that both resultant linear and rotational accelerations should be reported with head impact metrics.[110] As there is an improved correlation between impact biomechanics and the occurrence of a concussion, than when linear accelerations are reported alone.[12] Research[18, 111-114] suggests that the brain is more sensitive to rotational than linear accelerations. Rotational accelerations are reported[12, 115] to be correlated to the strain response of the brain and the primary mechanism for diffuse brain injury including concussion, contusion, axonal injuries and loss of consciousness.[111, 112, 116, 117] Linear accelerations are reported to result in the intracranial pressure response of the brain and be the primary mechanism for skull fractures and epidural haematomas.[115, 118] Reporting both linear and rotational accelerations should assist with identification of possible brain injury.

More recently[57, 58] resultant linear and rotational acceleration results have been combined into a risk weighted exposure (RWE) metric. This metric can be beneficial for fully capturing the linear (RWE_{Linear}), rotational ($RWE_{Rotational}$) and combined probability (from linear and rotational) (RWE_{CP}) of the risk of a concussion as it accounts for the frequency and severity of each player's impacts. The HIC and GSI are the most frequently utilised head injury assessment functions in helmet and traffic restraint safety standards,[12, 119] however this was not reflected in the sport head impact studies reviewed. Based on the Wayne State University tolerance curve,[94] the HIC and GSI criteria are considered plausible ways of determining relative risk of severe head injury[120] but they do not account for the complex motion of the brain, or the contribution of resultant rotational acceleration to the head.[12, 14, 101] In particular the HIC only deals with frontal impacts and was not designed to be used for lateral impacts that can be found in head impact biomechanics[119] and arbitrarily defines an 'unsafe pulse" within a 'safe pulse" by discounting any data outside the two time points chosen for the calculation of the HIC value.[121] The GSI and HIC may be beneficial for evaluating acute head trauma due to single impacts but they are reportedly not beneficial for repeated impacts at lower acceleration magnitudes[119] such as those found in contact sports such as American football, rugby union and soccer. The inclusion of the HIC and GSI by studies reporting on head impact biomechanics may be more historical thus providing the ability for inter-study comparisons with previous studies. However, as they are used to calculate multiple impacts and provide a nonsensical number, the value of these metrics are limited. The use of HIC and GSI in future studies, and the value that these metrics provide, needs to be standardised. Consensus is required on the incorporation of these and other biomechanical metrics into future research.

4.4 Limitations in the use of accelerometery

The use of accelerometers to record and assess movement is not new to the scientific community.[122, 123] There have been some inter-study and international comparability limitations reported for use of accelerometers to report physical activity.[62] The identified limitations for physical activity accelerometers may be identical to areas now being faced by studies reporting the biomechanics of impacts to the head. The majority of studies reporting head impact biomechanics have utilised HITS,[20-24, 32, 37, 42, 49, 51-54, 57, 58, 61, 65-83, 85, 86, 88-91] or a variant.[26] More recently, an electronic mouthguard has been used to assess head impacts in rugby union.[27]

The issues identified with the use of accelerometers for physical activity[62] include affordability of the accelerometers,[62] and the administration burden[62] to the participants and researcher(s) given post data collection analysis. The choice of accelerator brand,[124] generation[125] and firmware version,[126] wearing position[127] based on the sports code requirements (i.e. helmet mounted vs. headband mounted vs. mouthguard embedded vs. patch), specifics of the research being undertaken such as the epoch length[128, 129] (match vs. training vs. combined), data imputation methods,[130] dealing with spurious data[131] and the reintegration of smaller epochs into larger epochs[132] are all considerations for use of accelerometers. In addition to the issues identified, there are technological developments, emerging methodological questions and a lack of academic consensus that may also hinder the development of uniformity in the utilisation of accelerometers[62] for recording head impact biomechanics.

In comparing the New Zealand rugby union data with data collected with the use of the HITS, it must be noted that these are different impact telemetry systems. The mouthguard is reported to have a 10% error for linear and rotation acceleration and for angular velocity with an average offset of 2° for azimuth and elevation impact location.[133, 134] Although the correlation of the AIM mouthguard with laboratory head-forms is good, the impact measurements should be assumed to have some form of error that is dependent on impact conditions and the measure of interest and the variability tested.[89, 135] It is unlikely that the mouthguard was tested under all of the activities seen in rugby union matches such as the rucks, mauls, lineouts and scrum situations. How these rugby activities correlate to the laboratory conditions is unknown. .Although the majority of the impact biomechanics studies reported in this review are helmet based telemetry systems, there is a paucity of studies reporting on head impact biomechanics with other systems such as the mouthguard and headband. In addition there are no published studies comparing the HITS with other forms of impact telemetry systems such as the X2Biosystems All-In-Mouth (AIM) mouthguard.

A final consideration to the use of accelerometers in recording impacts is the need for concurrent video-analysis to enable comparison and verification of the impacts. This would enable the identification of non-impact activities where an impact has been recorded such as post-try celebrations, dropping equipment onto the ground, or other activities where the equipment may record an impact. In the case of the New Zealand rugby union data, only impacts that occurred in the tackle with the player standing were able to be verified.[27] The percentage of impacts that were identified at the 10*g* inclusion limit, that were able to be visualised by video review and analysis, varied from 65% to 85% of the total impacts recorded per match.[27]

4.5 What are the long term implications of repeated head impacts?

The use of impact tolerance and impact severity level data may be important if a risk assessment is undertaken for possible long term implications from repetitive head impacts (RHI). Recently in a small sample[66] of collegiate players with no reported concussions after a season of American football, there were white matter changes that correlated with multiple head impact measures. Participants with more than 30-40 RHI's with peak rotational accelerations >4,500 radians per second per second (rad/s^2) per season (r=0.91; p<0.001), and more than 10-15 RHI's >6,000 rad/s^2 (r=0.81; p<0.001), were significantly correlated with post-season white matter changes.[66] These changes post season imply a relationship between the number of RHIs that occur over a season of American football and white matter injury, despite no clinically evident concussion being recorded.[66]

The inclusion of impact tolerances and impact severity levels may assist with the identification of players at risk of possible long term injuries. Impact tolerance may also act as an indicator of when to rest players if they are exposed to RHIs above >4,500 rad/s^2 and >6,000 rad/s^2. This type of information will assist in formulating a detailed understanding of the exposure and mechanism of injury of concussion.[53, 136] Further research is required to evaluate the injury tolerance of concussive type injuries, to develop interventions to reduce the likelihood of any concussive type injuries, and to develop exposure durations and stand down periods to establish a broader understanding of the potential role of subconcussive events and long term health.[53]

5. Conclusion

This study identified the methodological differences in the threshold limits of impacts to the head as a result of participation in contact sports. Of the 34 studies, 39% reported impacts at the 10*g* impact threshold while 22% of studies used the 14.4*g* impact threshold. Resultant linear accelerations were most frequently reported (91%) while 76% reported resultant rotational accelerations. Nearly three-quarters (74%) of studies reported both resultant linear and rotational accelerations. Impact data were most frequently (52%) reported as mean ± standard deviation (±SD). Some (10%) studies reported the head impact data as median, but not all (4%) included the interquartile ranges (IQR) for these data.

The influence of head impact thresholds was shown using head impact data obtained from 38 senior amateur rugby union players during 19 matches in New Zealand. Application of the varied impact thresholds resulted in 20,687 impacts >10*g*; 11,459 (44.6% less), impacts >15*g*; and 4,024 (80.5% less) impacts >30g.

Given head impacts are likely to cause concussive injury, and the number of head impacts may be related to a potential concussion threshold (i.e. a cumulative threshold), the number and severity of head impacts should be monitored in players. However, impacts can be measured with several technologies (e.g. instrumented behind the ear patches, mouthguards, head gear), and thresholds have differed for reporting impact data in contact and collision sports. Consensus is therefore required to identify the reporting modalities (e.g. linear threshold, descriptive calculations), utilised in future impact studies to enable between study comparisons. Until in-field validation studies are completed, it is recommended that data should be reported as mean ± standard deviation, median and interquartile ranges using the 10*g* impact threshold.

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TABLES

Table 1: MOOSE scores, data acquisition impact thresholds, study groups, sporting codes and duration, instrumented equipment, participant numbers, impacts recorded for total and per player, and metrics for reporting data.

Table 2: Differences in the resultant linear (PLA(*g*)) and rotational (PRA(rad/s^2)) accelerations, head impact criterion (15ms) (HIC_{15}) and Gadd severity index (GSI) at different impact thresholds by the mean and standard deviation (±SD), median [25^{th} to 75^{th} percentile] and 95^{th} percentile for senior amateur rugby union players.

FIGURES:

Figure 1: Flow of identification, screening, eligibility and study inclusion of previously published studies.

Figure 2: Percentage of impacts removed when applying different data impact threshold limits compared with original 10*g* threshold limit for the New Zealand data set of head impacts to senior amateur rugby union players for one season.

Table 1: MOOSE scores, data acquisition impact thresholds, study groups, sporting codes and duration, instrumented equipment, participant numbers, impacts recorded for total and per player and metrics for reporting data

Study	MOOSE[63] Score	Data acquisition Limit (*g*)	Study group	Sport, No. seasons	No Participants	Impacts total	Impacts per player	Raw data		Derived variables				Reporting statistics				
								PLA(*g*)	PRA(rad/s^2)	HIC_{15}	HIC_{36}	GSI	HITsp	Mean (SD)	Median	IQR	95%	Other
Hernandez et al.[64]	4/6 (67%)	7; 10	Coll; P	Am F, MM, B			2 concussions; T: 513	Y	Y						Y	Y		
Brolinson et al.[65]	5/6 (83%)	10	Coll	Am F, 2	52	11,604	T; 223[a]	Y						Y				
Bazarian et al.[66]	5/6 (83%)	10	Coll	Am F, 1	10	9,769	T; 977[a]	Y	Y					Y				
Crisco et al.[21]	5/6 (83%)	10	Coll	Am F, 3	314	286,636	T: 420;[b] P: 250;[b] M: 128[b]	Y	Y				Y		Y		Y	Y
Crisco et al.[67]	5/6 (83%)	10	Coll	Am F, 2	254	184,358	T: 726[a]	Y	Y				Y				y	
Daniel et al.[54]	5/6 (83%)	10	Youth	Am F, 1	7	748	T: 107; P: 63; M: 44	Y	Y						Y		Y	Y
Duma et al.[49]	5/6 (83%)	10	Coll	Am F, 1	38	3,312	T: 87[a]	Y		Y		Y		Y				
Funk et al.[68]	5/6 (83%)	10	Coll	Am F, 4	98	37,128	T: 379[a]	Y		Y								
Hanlon et al.[26]	5/6 (83%)	10[H]	Youth	Soccer, P	24	47 H 20 NH	N/S	Y	Y	Y		Y						
Harpham et al.[69]	5/6 (83%)	10	Coll	Am F, 1	38	N/S	N/S	Y	Y				Y	Y				
King et al.[27]	5/6 (83%)	10[M]	Snr Amat	RU, 1	38	20,687	T: 564; M: 77	Y	Y					Y				
Mihalik et al.[61]	5/6 (83%)	10	Coll	Am F, 2	72	57,024	T: 9,504[a]	Y						Y				Y
Mihalik et al.[70]	5/6 (83%)	10	Youth	IH, 1	37	7,770	T: 1,945[a]	Y	Y					Y				Y
Mihalik et al.[71]	5/6 (83%)	10	Youth	IH, 2	52	12,253	T: 223;[b] P: 83;[b] M: 24[b]	Y	Y				Y	Y			Y	Y
Mihalik et al. [72]	5/6 (83%)	10	Youth	IH, 1	16	4,608	T: 288[a]	Y	Y				Y	Y				Y
Munce et al.[73]	5/6 (83%)	10	Youth	Am F, 1	22	6,183	T: 281[a]	Y	Y				Y	Y	Y		Y	
Ocwieja et al.[74]	5/6 (83%)	10	Coll	Am F, 1	46	7,992	T: 174[a]	Y	Y				Y	Y				Y
Reed et al.[24]	5/6 (83%)	10	Youth	IH, 1	13	1,821	T: 140; M: 5	Y	Y	Y	Y							Y
Rowson et al.[22]	5/6 (83%)	10	Coll	Am F, 1	10	1,712	T: 171[a]	Y	Y					Y				Y
Schnebel et al.[52]	5/6 (83%)	10	Coll	Am F, 1	40	54,154	T: 1,354[a]	Y										Y
		10	HS	Am F, 1	16	8,326	T: 520[a]	Y										Y
Beckwith et al.[75, 76]	5/6 (83%)	14.4	Coll / HS	Am F, 6	95	161,732	T: 1,702[a]	Y	Y	Y	Y							Y
Broglio et al.[77]	5/6 (83%)	14.4	HS	Am F, 1	42	32,510	T: 744; P: 11;[c] M: 24	Y	Y				Y					
Cobb et al.[20]	5/6 (83%)	14.4	Youth	Am F, 1	50	11,978	T: 240; P: 10; M: 11	Y	Y						Y		Y	Y
Crisco et al.[53]	5/6 (83%)	14.4	Coll	Am F, 1	188	3,878	T: 21;[a] P: 6; M: 14											Y
Daniel et al.[78]	5/6 (83%)	14.4	Youth	Am F, 1	17	4,678	T: 275;[a] P: 163; M: 112	Y	Y						Y		Y	Y
Rowson et al.[51]	5/6 (83%)	14.4	Coll	Am F, 2	314	300,977	T: 959[a]		Y									
Talavage et al.[42]	5/6 (83%)	14.4	HS	Am F, 1	21	15,264	T: 727[a]											
Urban et al.[58]	5/6 (83%)	14.4	HS	Am F, 1	40	16,502	T: 413[a]	Y	Y					Y	Y	Y	Y	
Young et al.[23]	5/6 (83%)	14.4	Youth	Am F, 1	19	3,059	T: 161; P: 95; M: 65	Y	Y					Y	Y		Y	Y
Broglio et al.[79]	5/6 (83%)	15	HS	Am F, 3	78	54,247	T: 695[a]	Y	Y					Y				Y
Broglio et al.[80]	5/6 (83%)	15	HS	Am F, 1	35	19,224	T: 549;[a] P: 9; M: 25	Y	Y									Y
Broglio et al.[37, 81]	5/6 (83%)	15	HS	Am F, 4	95	101,994	T: 652	Y	Y				Y	Y				Y
Eckner et al.[82]	5/6 (83%)	15	HS	Am F, 2	20	30,298	T: 1,515[a]	Y	Y				Y	Y	Y	Y	Y	Y
Martini et al.[83]	5/6 (83%)	15	HS	Am F, 2	83	35,620	T: 429[a]	Y	Y				Y	Y				
Wilcox et al.[25]	5/6 (83%)	20	Coll	IH M/F,3	91	37,411	T: 19,980[d] / 17,531[e]	Y	Y				Y		Y	Y	Y	Y
Wilcox et al.[84]	5/6 (83%)	20	Coll	IH M/F,1/3	54	616	T: 270[d] / 242	Y	Y				Y	Y				
Wong et al.[85]	5/6 (83%)	30	Youth	Am F, 1	22	480	T: 22;[a] P: 4; M: 2	Y						Y				
Gysland et al.[32]	5/6 (83%)	<60 >90	Coll	Am F, 1	46	N/S	T: 1,177; 12 >90*g*	Y										Y
McCaffrey et al.[86]	5/6 (83%)	<60 >90	Coll	Am F, 1	43	N/S	N/S	Y						Y				Y
Fréchède et al.[10] [A]	4/6 (67%)	Recon	Prof	AFL / RU, 3	-	-	N/S		Y	Y				Y				Y
McIntosh et al.[87] [A]	4/6 (67%)	Recon	Prof	AFL, 3	-	-	N/S	Y	Y									Y
Pellman et al.[16] [A]	4/6 (67%)	Recon	Prof	Am F, 5	-	-	N/S	Y	Y	Y				Y				
Zhang et al.[12] [A]	4/6 (67%)	Recon	Lab	-	-	-	-	Y	Y	Y				Y				Y
Breedlove et al.[88] [A]	4/6 (67%)	N/S	HS	Am F, 2	24	N/S	N/S	Y	Y						Y			Y
Duhaime et al.[89] [A]	4/6 (67%)	N/S	Coll	Am F, IH, 4	450	486,594	T: 1,081[a]	Y	Y									
Greenwald et al.[90] [A]	4/6 (67%)	N/S	Coll / HS	Am F, 3	449		17 concussions only	Y	Y	Y			Y				Y	
Guskiewicz et al.[91] [A]	4/6 (67%)	N/S	Coll	Am F, 2	88	104,714	T: 1,190[a]	Y						Y				
Rowson et al.[57] [A]	4/6 (67%)	N/S	Coll	Am F	N/S	63,011	Combined data	Y	Y									Y
Wilcox et al.[92]	4/6 (67%)	N/S	Coll	IH F, 3	58		9 concussions	Y	Y				Y	Y				
Mean study quality	4.8 ±0.4 (79.6% ±7.0)						Percentage of studies	91.5	92.0	76.6	76.0	21.3	18.0	4.3	4.0	6.4	4.0	

Instrumented equipment used is helmet unless the data acquisition limit is reconstructed = Recon, or superscript M = Mouthguard or H = Headband. Coll = Collegiate; HS = High School; Snr Amat = Senior Amateur; Prof = Professional; Am F = American Football; IH = Ice Hockey; RU = Rugby Union; AFL = Australian Football League; MM = Mixed martial Arts; B = Boxing; T = Total impacts; P = Practice Impacts; M = Match impacts; a = calculated number of impacts; b = Median results; c = contact practice; d = male; e = female; H = Header;

NH = non-header; N/S = Not Stated; PLA(g) = Peak Linear Acceleration; PRA(rad/s2) = Peak Rotational Accelerations in radians/second/second (rad/s^2); HIC_{15} = Head Impact Criterion 15 milliseconds; HIC_{36} = Head Impact Criterion 36 milliseconds; GSI = Gadd Severity Index; HITsp = Head Impact Telemetry severity profile; SD = Standard Deviation; IQR = Inter-Quartile Range; 95% = 95^{th} Percentile.

Table 2: Differences in the resultant linear (PLA(*g*)) and rotational (PRA(rad/s^2)) accelerations, head impact criterion (15ms) (HIC_{15}) and Gadd severity index (GSI) at different impact threshold limits by the mean and standard deviation (±SD), median [25th to 75th percentile] and 95th percentile for the New Zealand senior amateur rugby union total player dataset.

Data acquisition impact threshold (*g*)	No of impacts	Resultant Linear Accelerations (PLA(*g*))			Resultant Rotational Accelerations (PRA(rad/s^2))			Head Impact Criterion 15ms (HIC_{15})			Gadd Severity Index (GSI)		
		Mean ±SD	Median [25th-75th]	95%	Mean ±SD	Median [25th-75th]	95%	Mean ±SD	Median [25th-75th]	95%	Mean ±SD	Median [25th-75th]	95%
10	20,687	22 ±16	16 [12-26]	53	3,903 ±3,949	2,625 [1,324-4,934]	12,204	32 ±99	9 [5-25]	128	48 ±118	15 [8-398]	192
11	17,747	24 ±17	18 [13-29]	56	4,255 ±4,096	2,898 [1,549-5,389]	12,945	37 ±106	11 [6-30]	145	55 ±126	19 [10-47]	218
12	15,454	26 ±17	20 [15-31]	59	4,603 ±4,214	3,181 [1,781-5,860]	13,581	42 ±112	14 [7-35]	160	62 ±134	23 [12-55]	241
13	13,825	28 ±17	22 [16-32]	62	4,858 ±4,293	3,423 [1,967-6,263]	13,948	46 ±118	17 [9-40]	176	69 ±140	27 [14-62]	262
14	12,531	29 ±18	24 [18-34]	64	5,079 ±4,368	3,589 [2,123-6,596]	14,325	51 ±123	19 [10-44]	188	75 ±146	31 [17-69]	278
15	11,459	31 ±18	25 [19-35]	65	5,286 ±4,438	3,774 [2,263-6,908]	14,647	55 ±128	22 [12-49]	205	80 ±151	34 [19-76]	297
16	10,570	32 ±18	26 [20-36]	67	5,478 ±4,510	3,936 [2,400-7,180]	14,994	59 ±133	24 [14-53]	215	86 ±156	38 [21-82]	318
17	9,784	33 ±18	27 [21-38]	68	5,655 ±4,565	4,082 [2,538-7,394]	15,235	63 ±137	27 [15-57]	228	92 ±161	41 [24-88]	331
18	9,095	34 ±18	28 [22-39]	70	5,799 ±4,610	4,173 [2,644-7,567]	15,486	67 ±141	29 [17-62]	241	97 ±165	45 [27-95]	348
19	8,500	35 ±19	29 [23-40]	71	5,939 ±4,662	4,265 [2,731-7,744]	15,823	70 ±145	32 [18-66]	253	103 ±169	49 [29-102]	364
20	7,934	36 ±19	30 [24-41]	74	6,072 ±4,716	4,357 [2,810-7,931]	16,256	74 ±150	34 [20-70]	263	109 ±174	53 [31-109]	374
21	7,430	37 ±19	31 [25-42]	76	6,206 ±4,757	4,483 [2,896-8,158]	16,470	79 ±154	37 [22-75]	275	115 ±178	57 [34-114]	391
22	6,938	39 ±19	32 [26-44]	77	6,363 ±4,801	4,595 [2,992-8,426]	16,806	83 ±158	40 [24-80]	291	121 ±183	62 [37-121]	415
23	6,463	40 ±19	33 [27-45]	80	6,519 ±4,859	4,722 [3,096-8,628]	17,073	88 ±163	43 [26-85]	302	127 ±188	67 [40-129]	444
24	6,060	41 ±19	34 [28-46]	82	6,656 ±4,906	4,835 [3,201-8,798]	17,282	92 ±167	46 [28-90]	318	134±192	71 [43-135]	466
25	5,666	42 ±20	35 [29-47]	83	6,819 ±4,952	4,965 [3,305-9,012]	17,435	97 ±172	49 [31-95]	337	141 ±197	76 [47-144]	485
26	5,275	43 ±20	36 [30-48]	84	6,977 ±4,986	5,101 [3,428-9,297]	17,622	102 ±177	53 [33-101]	357	148 ±202	81 [50-152]	512
27	4,955	44 ±20	37 [31-49]	87	7,107 ±5,036	5,210 [3,495-9,459]	17,844	107 ±181	57 [35-107]	389	155 ±206	86 [54-162]	536
28	4,642	45 ±20	39 [32-51]	88	7,261 ±5,079	5,339 [3,607-9,704]	18,131	113 ±186	60 [38-114]	396	163 ±211	93 [58-173]	557
29	4,305	47 ±20	40 [33-52]	91	7,448 ±5,130	5,492 [3,778-9,917]	18,221	119 ±192	65 [41-123]	407	172 ±217	99 [64-186]	583
30	4,024	48 ±20	41 [34-54]	92	7,597 ±5,187	5,624 [3,875-10,129]	18,436	125 ±197	69 [44-131]	420	180±221	106 [68-196]	606

PLA (g) = peak linear acceleration; PRA (rad/s^2) = peak rotational acceleration in radians/second/second (rad/s^2); Significant difference (p<0.05) than: (a) = 10*g*.

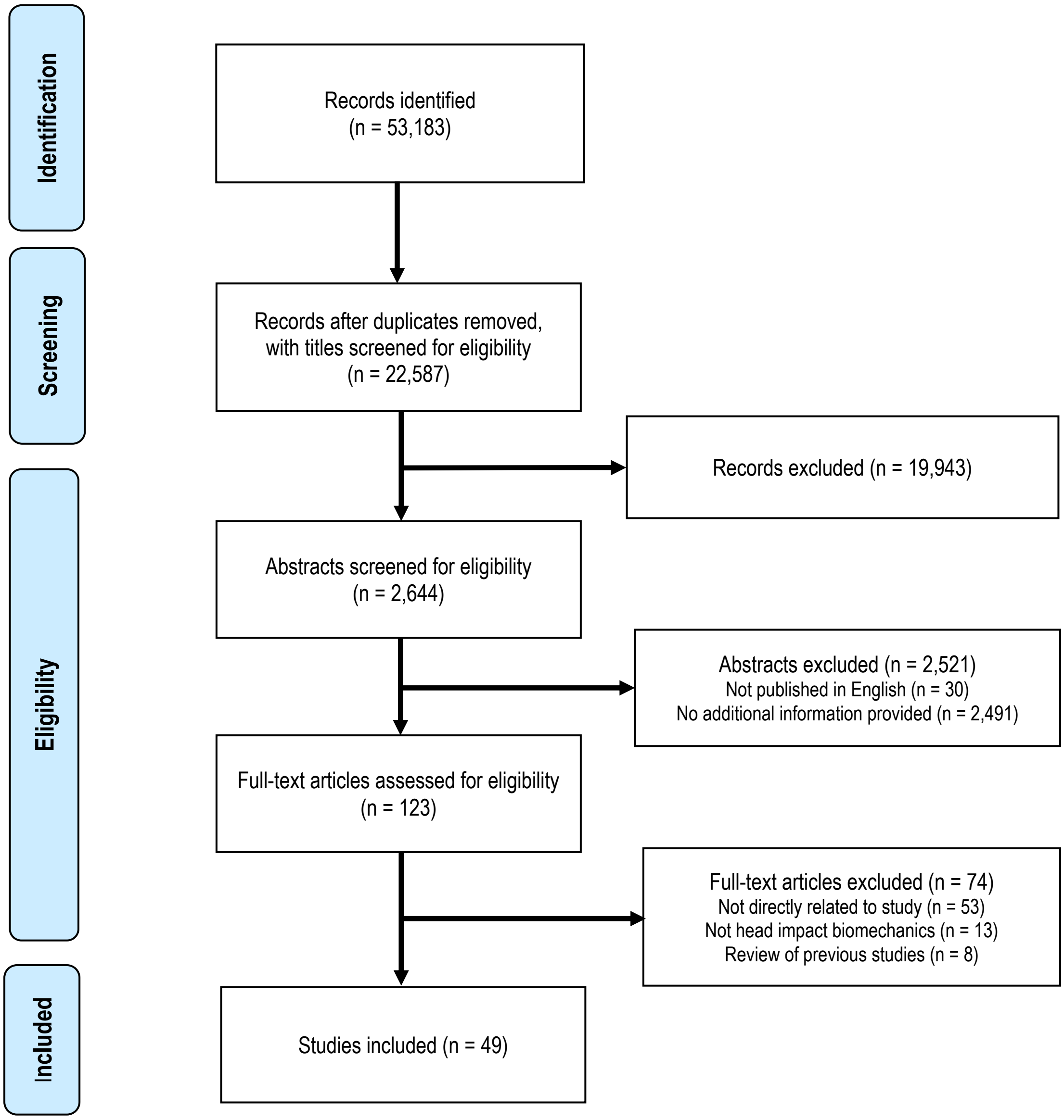

Figure 1: Flow of identification, screening, eligibility and study inclusion of published studies.

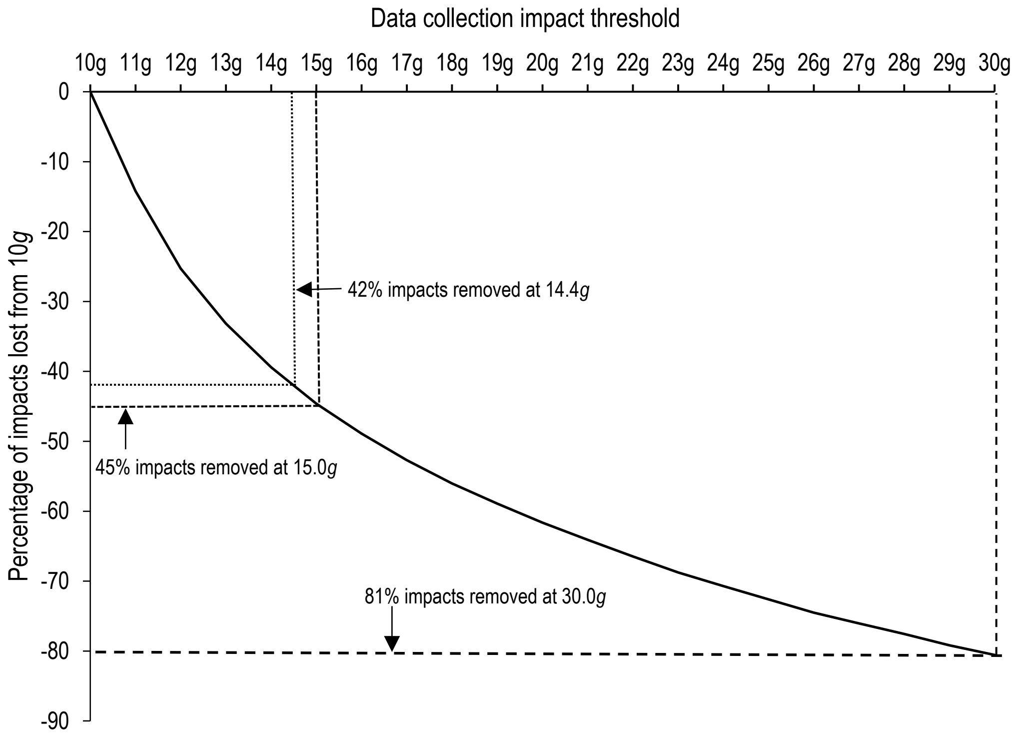

Figure 2: Percentage of impacts removed when applying different data impact threshold limits compared with original 10*g* threshold limit for the New Zealand data set of head impacts to senior amateur rugby union players for one season.