Head Impact Sensors:
An Examination of their Performance, Validity and Utility in Boxing Sparring

Enora Le Flao

A thesis submitted to Auckland University of Technology in fulfilment of the requirements of the degree of Doctor of Philosophy (PhD)

July 2021

Primary Supervisor: Dr Robert Borotkanics

Co-Supervisor: Dr Gunter Siegmund

Co-Supervisor: Dr Nigel Harris

Advisory Committee Member: Dr Seth Lenetsky

Sports Performance Research Institute New Zealand

School of Sport and Recreation

Faculty of Health and Environmental Sciences

Auckland University of Technology, Auckland, New Zealand

Abstract

Concussions represent a major challenge for sports medicine, as they are frequent and may lead to long-term health impairments. To better understand concussions and the risks associated with participation in contact sports, wearable technology has been developed and used with increasing frequency. This technology aims to measure head kinematics upon impact, based on the assumption that the sudden change of head velocity is related to brain damage. While understanding this relationship would allow the development of injury prevention strategies, it has remained elusive, and issues relative to the validity and accuracy of the sensors when used *in-vivo* may have contributed to this apparent lack of association.

The current thesis aimed to assess the performance of head impact sensors in the laboratory and *in-vivo,* and improve our understanding of how these sensors can be soundly used. Specific aims were to:

1. Appraise via literature review the achievements and limitations of head impact research.
2. Assess the validity of a select instrumented mouthguard under controlled impacts in the laboratory.
3. Evaluate and compare the performance and validity of an instrumented mouthguard, a skin patch and a patch attached to the headgear during boxing sparring.
4. Explore the feasibility of individual-specific approaches to associate head impact kinematics with acute concussion symptoms resulting from sparring.

Our findings highlighted that head impact sensors have been used to better understand the consequences of head impacts at various levels: from a single concussive impact to a lifetime of contact sports participation. However, the progression of head impact research is limited by large variability in technologies and methodologies, and by the inclusion of false positive impacts. The laboratory study found that the instrumented mouthguard under evaluation measured linear acceleration and angular velocity signals that adequately represented headform motion when the mouthguard was tightly coupled to the headform, although the accuracy was moderate. The *in-vivo* study highlighted several limitations for the use of the instrumented mouthguard, skin patch, and headgear patch during boxing sparring:

- Large proportions of events recorded by the sensors (50-80%) did not meet a set of pre-defined quality criteria: the kinematic signals suggested the sensors moved independently of the skull. Such events were associated with higher peak accelerations that might be spurious.

- Impacts landing close to the sensors were associated with more events being triggered and more events suggesting skull/sensor decoupling, for all three sensors.
- The skin and headgear patches recorded twice as many video-verified acceleration events as the mouthguard, and there was little to no association in peak accelerations between the patches and the mouthguard.

We also observed that some boxers regularly experience concussion symptoms from sparring, and our exploration suggested that individual-specific approaches are beneficial to understanding the association between head impact kinematics and symptoms. Due to our small sample size, there was limited evidence that the accumulation of head impacts or the highest-magnitude impact sustained during a session were associated with changes in self-reported symptoms.

Based on our findings, we recommend that future researchers systematically assess the quality of the raw kinematic signals as part of the data cleaning process. We also recommend that further work be done to explore the use of individualized approaches to better understand the association between head impact exposure and signs and symptoms of concussion. We reason that our findings and recommendations will advance research and sound application of head impact technology going forward.

Table of contents

List of figures

List of tables

Attestation of authorship

I hereby declare that this submission is my own work and that, to the best of my knowledge and belief, it contains no material previously published or written by another person (except where explicitly defined in the acknowledgements), nor material which to a substantial extent has been submitted for the award of any other degree or diploma of a university or other institution of higher learning.

Enora Le Flao
10 July 2021

Co-authored works

The current thesis contains jointly authored articles published, submitted or to be submitted to peer-reviewed journals, and conference presentations. The following manuscripts and publications are central to the thesis work and contribution reflected herein.

Work submitted for publication

Chapter	Title	Authors' contribution	Publication plan
Chapter 2	Benefits and limitations of *in-vivo* head impact monitoring research in sports: A systematic review	Enora Le Flao[1] (90%), Gunter Siegmund[2] (5%), Robert Borotkanics[3] (5%)	Under review in Sports Medicine
Chapter 3	Laboratory validity of the CSx instrumented mouthguard for measuring head impact biomechanics	Enora Le Flao[1] (80%), Khyati Verma[4] (14%), Nicolas Bourdet[4] (1%), Gunter Siegmund[2] (2%), Remy Willinger[4] (1%), Robert Borotkanics[3] (2%)	Submitted in February 2021 to Annals of Biomedical Engineering

The following chapters are in preparation for publication:

Chapter	Title	Authors' contribution
Chapter 4	Performance of head impact sensors to record video-verified head impacts in boxing sparring	Enora Le Flao[1] (90%), Robert Borotkanics[3] (4%), Gunter Siegmund[2] (3%), Seth Lenetsky[1,5] (3%)
Chapter 5	Head impact kinematics during boxing sparring: assessment and comparison of an instrumented mouthguard, a skin patch, and a headgear patch	Enora Le Flao[1] (90%), Robert Borotkanics[3] (4%), Gunter Siegmund[2] (4%), Seth Lenetsky[1,5] (2%)
Chapter 6	Individual-specific associations between head impact kinematics and acute symptoms changes resulting from boxing sparring: An exploratory study	Enora Le Flao[1] (90%), Robert Borotkanics[3] (4%), Gunter Siegmund[2] (3%), Seth Lenetsky[1,5] (3%)

We, the undersigned, hereby agree to the percentages of participation to the works identified above:

Supervisory Committee

Robert Borotkanics[3]
Primary Supervisor

Gunter Siegmund[2]
Secondary Supervisor

Seth Lenetsky[1,5]
Advisor

Collaborators

Khyati Verma[4]

Nicolas Bourdet[4]

Remy Willinger[4]

[1] Sports Performance Research Institute New Zealand (SPRINZ), Faculty of Health and Environmental Science, Auckland University of Technology, Auckland, New Zealand

[2] MEA Forensic Engineers & Scientists, Richmond, Canada

[3] Faculty of Health and Environmental Science, Auckland University of Technology, Auckland, New Zealand

[4] Laboratoire des Sciences de l'Ingénieur, de l'Informatique et de l'Imagerie, Université de Strasbourg & CNRS-UMR 7357, Strasbourg, France

[5] Canadian Sports Institute Pacific, Victoria, Canada

Other contributions

In addition to peer-reviewed manuscripts/publications, I wish to acknowledge the contributions of the following individuals for their valuable mentorship, tangential works or collaborations: Matt Brughelli[1], Patria Hume[1,6], Doug King[1,7], Michael Hamlin[8], Sohei Takamori[8].

Section	Title	Authors' contribution	Publication plan
Appendix G	Head impact monitoring: what new methodologies could do for concussion biomechanics	Enora Le Flao[1] (90%), Patria Hume[1,6] (7%), Doug King[1,7] (3%)	Presented at the 36th conference of the International Society of Biomechanics in Sports (2018) and published in the proceedings
Appendix H	Validation of the linear acceleration measured by instrumented mouthguards for *in-vivo* head impact monitoring.	Enora Le Flao[1] (80%), Khyati Verma[4] (14%), Nicolas Bourdet[4] (1%), Doug King[1,7] (1%), Patria Hume[1,6] (1%), Remy Willinger[4] (1%), Robert Borotkanics[3] (1%), Michael Hamlin[8] (0.5%), Sohei Takamori[8] (0.5%)	Presented at the International Society of Biomechanics conference (2019) and published in the book of abstracts
Appendix I	Assessing head/neck dynamic response to head perturbation: A systematic review.	Enora Le Flao[1] (85%), Matt Brughelli[1] (7%), Patria Hume[1,6] (4%), Doug King[1,7] (4%)	Published in Sports Medicine

[6] National Institute for Stroke and Applied Neuroscience, Auckland University of Technology, Auckland, New Zealand

[7] School of Science and Technology, University of New England, Armidale, NSW, Australia

[8] Lincoln University, Christchurch, New Zealand

Acknowledgements

I expected the PhD to be tough. I started mine relatively late, at 28, and I had seen many of my friends go through hard times. I had had a few of these times myself in my previous jobs, so I thought I was prepared…

I was so wrong.

This PhD has been by far the most difficult thing I have ever undertaken. It pushed me to edges I had never reached before and made me experience a whole new range of feelings. Even so, not once have I regretted embarking on this journey, and never have I truly thought about giving up. I had started the journey for a variety of reasons, and these reasons kept me going. The saying "What doesn't kill you makes you stronger" resonates very strongly with me now, for I am much, much stronger than I was when I started this journey.

I owe a lot of that newly-found strength to my supervisors. Robert, I will be eternally grateful for your unwavering support throughout the storm. Our long conversations about all things PhD-related, as well as things non-PhD related were what I needed. I honestly don't know if I would have been able to keep going, even less finish, without you. Thank you for joining this bumpy ride and for taking me to the finish line. Hopefully, we will meet again for a funnier kind of ride!

Gunter, thank you for your interest in my research and for your precious time. Your dedication to doing good research is inspiring. I deeply appreciate you putting the bar high (which is where it should be!) and I hope I met the expectations. Thank you for reminding me that "every day you learn something is a good day." I hope our paths will cross again.

Seth, your enthusiasm is contagious and has often (re-)motivated me! You have been advising me since before I even started this PhD, and I remember you saying you would have liked to supervise me. Even though the topic changed, we did it! Thank you for being part of the adventure. I am thrilled to have projects that will keep us working together in the future.

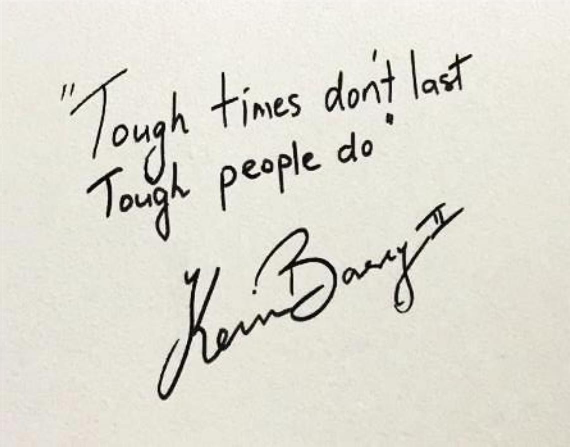
Robert, Gunter, Seth, thanks again for your invaluable guidance and for the push to do good and useful research. I feel extremely lucky to have had such a team! I am happy about the work we have done together and proud of this final document. I look forward to supervising someone one day and be able to pass on some of what you taught me.

I would also like to thank Nigel Harris for stepping up as mentor supervisor and allowing me to continue my PhD in the best conditions. To Matt Brughelli, Patria Hume and Doug King, who made me want to come to New Zealand in the first place and supervised me as I started this PhD: I have learned a lot from our interactions about research, academia, communication, and myself. I am definitely stronger from it, and I thank you for that. I have had the incredible opportunity to meet and collaborate with many people over the past five years. Khyati Verma, Remy Willinger, Nicolas Bourdet, thank you for the collaboration. Mike Hamlin and Sohei Takamori, thank you for allowing me a sneak peek of your data, this was incredibly helpful and helped me choose the right direction for this PhD. Liz Williams and Natalia Kabaliuk, thank you for offering your help and advice, I hope I will be able to return the favour one day. An enormous thank you to Cameron Todd and Alexis Pritchard for opening the doors of Wreck Room to me, and to all the boxers who participated. It was great to see that you were interested in this research and it motivates me to keep going!

There are so many friends I would like to thank, people who have supported me, whether they knew it or not, and people who were just there and made me smile… *merci, thank you, gracias, danke*. There are a few people that I'd like to thank specifically, whose friendship has been instrumental to my sanity and happiness. Edward Ashworth, you have been a rock during the toughest times and I can't thank you enough for being there and listening to me. I wished we had had even more amazing adventures and more occasions to sing in the car! Livvie Baudinet, Lauren Keaney, Alana Coombe and Jessica Bush: thank you girls for welcoming me into your lives, and for the endless discussions about how to take care of ourselves through the PhD struggles. You are strong and inspiring women and I can't wait to see more of what life brings you! Thank you to Steve Kolose (we made it despite everything!), Anja Zoellner, Dustin Oranchuk, Lesley Sommerfield, Aaron Uthoff, and many more! Thanks also to the fabulous SPRINZ/AUT staff, Jane Hall, Lesley Ferkins, Jacqui Pratt, Allan Carman, Luigi Bercades, JC Cronin, and more… your help or offer of help made a difference.

Finally, to my parents… the hardest thing in this whole journey was to see you suffer from my anxiety and not being able to fully enjoy our limited time together. I know you would have preferred if I had chosen a country that was not all the way across the world, but I also know you understood my choice. After all, you are the ones who raised me to be independent and pursue what I love! I cannot thank you enough for your unconditional support. I may not come back home, but I promise the next place I move to will also be worth the visit!

A mes parents… le plus dur pendant le cours de cette thèse a été de vous voir souffrir de mon anxiété, et de ne pas avoir été capable de profiter du si peu de temps que nous avions ensemble. Je sais que vous auriez préféré une destination plus proche de la maison, mais je sais aussi que vous comprenez les raisons qui m'ont poussées vers ce pays à l'autre bout du monde. Après tout, c'est vous qui m'avez inculqué cette indépendance, et ce désir de poursuivre mes rêves. Je n'ai pas les mots pour vous remercier de votre soutien et de votre aide pendant cette longue étape de ma vie. Je ne pense pas rentrer, mais je vous promets que la prochaine destination vaudra aussi le coup de la visite !



Writing on the wall of Wreck Room, where this PhD's core study took place.

Kevin Barry II represented New Zealand at the 1984 Los Angeles Olympics. "He fought bravely to make his way into the semi-finals, where he ran into the star of the American team, Evander Holyfield. They had a torrid fight and were both warned by the referee […] for fouls, Barry for holding behind the neck, Holyfield for hitting after the referee's call to "break". A few seconds before the end of the second round, Holyfield again hit on the break and the unprepared Barry was knocked to the canvas. He rose groggily and the fight was stopped. Then a sensation: Holyfield was disqualified. […] The result stood [despite American protests], but Barry was deprived of his chance to fight for the gold. He had been declared a knockout victim and therefore, under amateur boxing regulations, was not able to fight again for 28 days."
Source: https://www.olympic.org.nz/athletes/kevin-barry/

Ethics approval

Ethics approval to conduct this research was granted by the Auckland University of Technology Ethics Committee (AUTEC):

> **Head impacts monitored using mouth guards, patches and headgear: differences and similarities during sparring**. 20/153, accepted 5 August 2020 (Appendix C).

Grants and awards

The following competitive grants were obtained to support the research study “Head impact sensors: differences and similarities between mouthguards, patches and headgear sensors”, which resulted in Chapters 4, 5, and 6:

International Society of Biomechanics, International Travel Grant. 2020, US$2,500.

International Society of Biomechanics in Sport, Student Research Grant. 2020, €1000.

Chapter 1 – General introduction

1.1. Motivation for research

Concussions in sports have been recognised as a major issue in sports medicine, inciting calls for the ban of high-risk contact sports, such as boxing, mixed martial arts, or youth American football. However, it is also well established that participation in sport, including at the youth level, has multiple beneficial effects on long-term health. While increasing awareness around the short- and long-term consequences of concussions is crucial to allow athletes to make an informed choice, it is equally important to make these sports safer in general. With concepts of injury prevention in mind, wearable technology, namely head impact sensors, has been developed to measure head motion after an impact. One of the underlying ideas is that these sensors could help identify when an athlete has sustained too many, or too severe, impacts to the head. This technology could provide valuable information about aspects of participation that present high risks of concussion to the athletes, thereby motivating informed changes in rules and regulations. Additionally, surveillance applying this technology could ideally be used to assist with removal from play and clinical assessment.

However, there is a feeling that the field has grown too fast, aided by the popularity of these wearable sensors. Some basic requirements and necessary validations seem to have been neglected, and as a result, datasets are corrupted, compilation of study results is limited, and the relationship between head impacts and concussion remains elusive. With this PhD, we chose to take what could be seen as a step back, assessing and shedding light on some of the nuances associated with wearable head impact technology.

1.2. Background

Key terms

Concussion: in this doctoral work, we focus on the milder end of the traumatic brain injury spectrum, where concussions belong. We choose to use the definition of concussion provided by the Concussion in Sport Group in their 2017 consensus statement.[255, 256]

"Sport-related concussion [SRC] is a traumatic brain injury induced by biomechanical forces. Several common features that may be utilised in clinically defining the nature of a concussive head injury include:

- *SRC may be caused either by a direct blow to the head, face, neck or elsewhere on the body with an impulsive force transmitted to the head.*
- *SRC typically results in the rapid onset of short-lived impairment of neurological function that resolves spontaneously. However, in some cases, signs and symptoms evolve over a number of minutes to hours.*
- *SRC may result in neuropathological changes, but the acute clinical signs and symptoms largely reflect a functional disturbance rather than a structural injury and, as such, no abnormality is seen on standard structural neuroimaging studies.*
- *SRC results in a range of clinical signs and symptoms that may or may not involve loss of consciousness. Resolution of the clinical and cognitive features typically follows a sequential course. However, in some cases, symptoms may be prolonged.*

The clinical signs and symptoms cannot be explained by drug, alcohol, or medication use, other injuries (such as cervical injuries, peripheral vestibular dysfunction, etc) or other comorbidities (e.g., psychological factors or coexisting medical conditions)."

Exposure to head impacts hereby relates to the number, magnitude, and direction/location of head impacts sustained over a defined period, e.g., a game, a practice session or a full season of sports participation.[26, 91]

Concussion and its challenges

Concussion has become a worldwide concern in sports medicine.[96, 296] Leading reasons for this concern are the high incidence rate in various contact sports, such as American football, rugby, or combat sports,[31, 95, 132, 215, 231] and the risks of persistent deleterious effects on health.[247, 376] In addition to the immediate neurological dysfunction, concussed patients often show symptoms persisting past the typical 7-14 days window of recovery.[253, 256] Furthermore, athletes with a history of repetitive exposure to head impacts, with or without concussions,[280] can suffer from long-term impairments such as neurocognitive issues,[169, 171] memory deficits,[179] behavioural changes,[103, 140] headaches,[103] and altered movement coordination.[40, 259] Repetitive head impacts and previous concussions can also decrease an athlete's tolerance,[2, 374] showing the effect of accumulation and/or impaired recovery, and resulting in a progressive increase in injury risk.

To limit injury risks and the long-term effects of sports-related exposure to head impacts, the three pathways of injury prevention can be investigated.[113, 315]

1. *Primary prevention* intends to prevent the injury from occurring, usually through removal from participation or reduction of the factors contributing to injury.
2. *Secondary prevention* *"aims to detect the injury of interest at a point early enough in its development where intervention can prevent its progression or worsening."*[113]
3. *Tertiary prevention* aims to reduce complications associated with the injury, e.g., long-term consequences or risk of recurrent injury.

However, in the case of concussions, a challenge resides in that they are difficult to identify.[326] Particularly, a large part of the clinical diagnosis relies on the physician's assessment of the signs and symptoms experienced by the athlete.[256] However, it is common for athletes not to report symptoms, because they lack awareness about the potential consequences,[30, 79] or they do not want to be removed from play.[79, 254] Therefore, there is a need for objective and reliable tools to assist with the identification of concussed athletes, particularly in the detection of head impacts requiring clinical follow-up. Wearable head impact technology was developed to meet this need and has potential application, if soundly applied, to all levels of prevention.

Biomechanics for injury prevention

As defined above, the mechanics of concussion are accepted to be a direct impact to the head, face or neck or an impact elsewhere on the body that transmits an impulsive force to the head.[256] The forces affect the velocity of the skull, for example making it move from a stationary position (e.g., a punch to the head), or suddenly stopping it from moving (e.g., when the head hits the ground). The sudden change in velocity subjects the brain to deformations within the skull cavity.[142, 168] Specifically, linear motions (translation of the head) are related to intracranial pressure gradients that can cause focal injuries, while angular motions (rotation of the head) can lead to shear stresses and tensile strains within the brain's tissue.[202, 321] These two mechanisms can further generate elastic or permanent deformation of the various structures of the brain (e.g., circulatory system, axons, synapses, cell membranes).[33, 289] The resulting damage can then trigger a multi-faceted cascade of altered state and function including, but not limited to, ionic shifts and disturbances in cellular metabolism, neuroinflammation and disruption in neurotransmission, and increased permeability of the blood-brain barrier affecting the brain's lymphatic system.[146, 172, 342]

It is not currently possible or practical to measure pressure gradients, brain strains, tissue damage or neurometabolic disturbances during sports participation. However, it is relatively easy to measure the skull's change of velocity upon impact using accelerometers and angular rate sensors. Such sensors, namely head impacts sensors, have been used in various forms, starting with instrumented helmets in the mid-to-late 1900's,[1, 281, 314] and have become increasingly popular in the last 15 years. Head impact sensors measure the motion of the head on the assumption that brain strains and by extension, brain damage, are in some way proportional to the magnitude of this motion.[149, 281] *"The ultimate motivation for studying the biomechanics of brain trauma is to elucidate the relationship between kinematics of the entire intact skull and cellular-level impairment."*[289] Elucidating this relationship would potentially enable better injury management and prevention by:

- Preventing the injury (*primary prevention*): identifying, then reducing aspects of participation that are prone to head impacts and/or injury, e.g., implementing laws against body checking in youth ice hockey showed a 67% reduction in concussion risk.[122]
- Minimising the injury (*secondary prevention*): signalling when an athlete reaches a threshold in exposure and removing them from participation to prevent further insult. Research has shown that immediate removal after an injury is associated with shorter recovery.[74]
- Identifying injured athletes (*tertiary prevention*): raising awareness about an athlete who has been exposed to severe, and/or many head impacts during a session, to have him assessed by a physician. This would allow adequate identification of injury and subsequent clinical management.[133, 269, 286]

Attempts have been made to determine global concussion risk thresholds; for example, peak linear head accelerations (PLA) of 85 g and peak angular head accelerations (PAA) of 6,000 rad.s^{-2} have been proposed as injury thresholds.[148, 351, 423] A recent meta-analysis showed that the average values of peak head acceleration for concussed male athletes were 98.68 g (95 % confidence interval: 82.36 – 115.00) and 5,776.60 rad.s^{-2} (95 % CI 4,583.53 –6,969.67).[38] However, these proposed thresholds and average values should be interpreted with caution, as concussions in adult male athletes have been associated with impacts as low as 41 g and 3,000 rad.s^{-2}.[229, 250] On the opposite side of the spectrum, studies[261, 378] reporting PLA and PAA values as high as 180 g and 16,000 rad.s^{-2} did not report any diagnosed concussions. As a result of this variability, the relationship between skull kinematics and brain damage remains elusive.[269]

Thesis rationale

There are a variety of factors that could explain why the relationship between skull kinematics and brain damage or concussion has stayed out of reach. First, as mentioned above, the process of identifying a concussion is complex. Second, biomechanical forces might have unique effects and manifestations from one person to another because of differences in individual tolerance.[2, 39, 122, 149] Third, the association between repetitive subconcussive blows and the occurrence of injury remains unclear.[26, 27, 120, 409] There is currently no validated way to assess the effects of single vs. multiple events, nor to account for the time between events and potential recovery, or for the hypothetical decrease in tolerance. Fourth, the information available from a single head impact is complex, time-varying, and multi-dimensional, and has often been reduced to discrete metrics, resulting in an incomplete description of the kinematic patterns. Finally, the sensors may present important limitations related to the number and magnitude of head impacts recorded, particularly *in-vivo*. Several authors have reported errors in the number of impacts recorded,[86, 322] and the inclusion of invalid head acceleration events may pollute datasets in ways that have not been fully evaluated. Also, the validity of the kinematic measurements measured *in-vivo* may be limited,[367, 369, 418] particularly for angular kinematics, which are especially relevant to injury.[182, 212]

The rapid development of head impacts sensors, resulting from the increased awareness of concussions and their long-term consequences, highly publicised controversies involving professional American football,[189] and the rise of wearable technology globally, may have contributed to the limitations detailed above. For many sensors used in previous research, there has not been an exhaustive and rigorous assessment of their performance. This is particularly true for *in-vivo* use of the devices, where greater variability is typically observed due to the uncontrolled environment and to human factors and interactions. As a result, head impact datasets may contain erroneous data that is being used in biomechanical investigations, and that potentially hinder the search for the association between head kinematics and injury.

1.3. Research questions

The overriding objective of this thesis was to explore the performance and limitations of head impact sensors and improve our understanding of how these sensors can be soundly used *in-vivo*. A systematic literature review, a laboratory study and an *in-vivo* observational study were conducted to fulfil this objective. Specific research questions were developed within each chapter and focused around:

1. What did head impact research achieve with respect to injury prevention? What are the main limitations that the field faces?

2. How valid are the kinematic signals measured by an instrumented mouthguard under controlled impacts in the laboratory?

3. How do an instrumented mouthguard, a skin patch, and a headgear patch compare in terms of the number of head acceleration events recorded during boxing sparring?

4. Do an instrumented mouthguard, a skin patch, and a headgear patch measure the same head impact kinematics when comparing simultaneous video-verified head acceleration events?

5. Can head impact kinematics and concussion-related symptoms resulting from sparring be associated using individual-specific approaches?

Answering the questions outlined above would shed some light on the strengths and weaknesses of several types of sensors typically used in head impact research. By diving deep into both laboratory and *in-vivo* data, we hope to expand on the understanding of some of these weaknesses, and how they may affect the information that is reported and shared with the scientific community. This new knowledge may help appreciate why the association between head impact exposure and concussion has stayed out of reach for so long. The recommendations that will result from this doctoral work will be designed to guide future research and subsequently serve the development of injury prevention strategies.

1.4. Thesis structure

This thesis comprises five core chapters bounded by a general introduction (the current chapter) and a closing chapter presenting the key findings, general discussion, and conclusions (Chapter 7)(Figure 1.1). The first core chapter (Chapter 2) is a systematic review summarizing the research conducted using wearable head impact sensors during sports participation, highlighting findings and limitations. The limitations associated with the use of the sensors identified in the review motivated the development of the subsequent studies. Chapter 3 assesses the validity and accuracy of an instrumented mouthguard in the laboratory using time-series analyses. Chapters 4 and 5 investigate the *in-vivo* performance of three different sensors used simultaneously: an instrumented mouthguard, a skin patch, and a patch attached to the athlete's headgear. Boxing sparring was chosen for this investigation to provide an environment that optimised the ability to video-verify head impacts. Chapter 4 focuses on the number of impacts recorded by the three sensors, and Chapter 5 on the kinematic measurements. Finally, a subset of data collected during boxing sparring was used to examine the association between head impact kinematics and acute changes in self-reported symptoms (Chapter 6). This final core chapter was conducted in an exploratory way to assess the relevance of individual-specific analysis approaches.

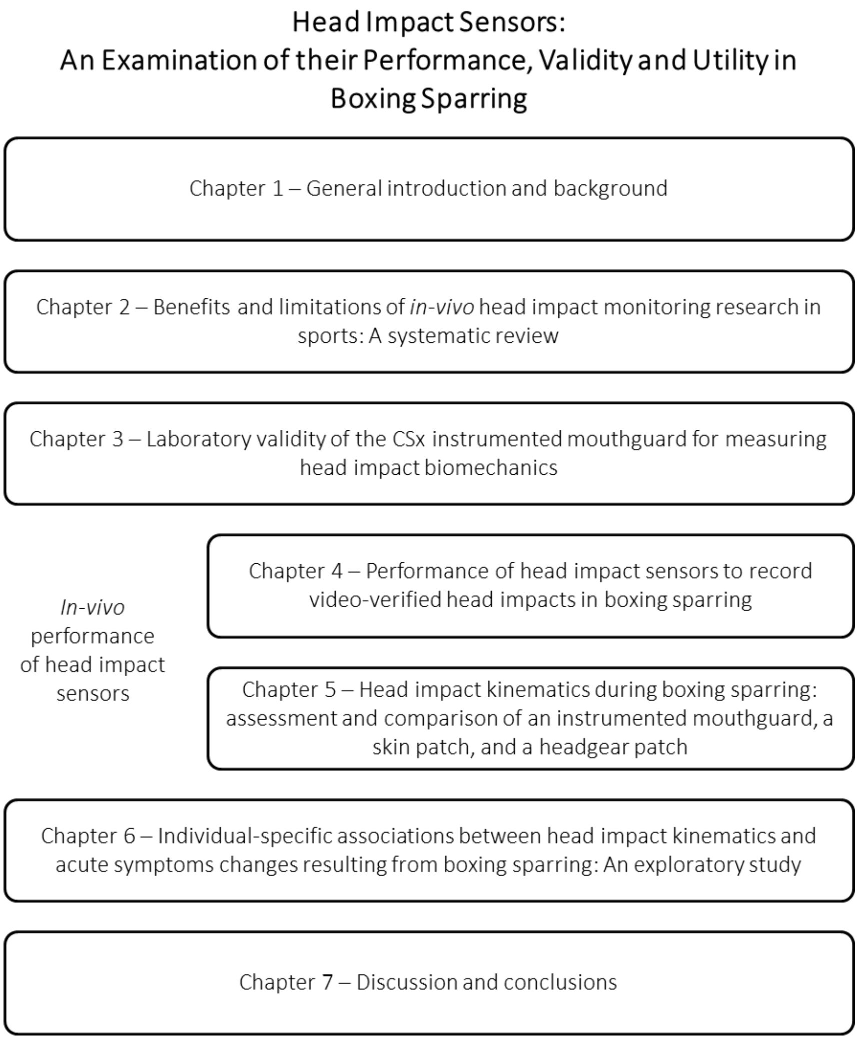


Figure 1.1. Overview of the thesis structure.

The authors' contribution and status of the manuscripts' acceptance can be found on page xiii. Each chapter is presented as a stand-alone chapter and, as such, may include repetition of information throughout the thesis.

Appendices A and B include supplementary materials for Chapters 2 and 3, respectively. Appendices C through F pertain to the *in-vivo* study and present the ethical approval documents, elements of protocol and supplementary materials. Specifically, Appendix F is a stand-alone analysis of the performance of the sensors' proprietary classification algorithms. Appendices G and H are conference abstracts associated with oral presentations given at international conferences. Appendix I is a systematic literature review on the methodologies used to assess neck neuromuscular performance, that was conducted before a change in scope for the PhD.

Chapter 2 – Benefits and limitations of *in-vivo* head impact monitoring research in sports: A systematic review

This chapter comprises the manuscript submitted for publication:

> Le Flao, E., Siegmund, G. P., & Borotkanics, R. J. Benefits and limitations of *in-vivo* head impact monitoring research in sports: A systematic review. (*Under review by Sports Medicine)*

Supplementary materials are presented in Appendix A. An electronic table, submitted as a spreadsheet, is temporarily available for viewing and/or download via the link: shorturl.at/msxCU

2.1. Abstract

Background: The number and magnitude of head impacts have been assessed in-vivo using inertial sensors to characterise the exposure in various sports and to help understand their potential relationship to concussion.

Objectives: To provide a comprehensive review of the field of in-vivo sensor acceleration events (SAEs) research in sport via the summary of data collection and processing methods, population demographics and factors contributing to athlete's exposure to SAEs.

Methods: The systematic search resulted in 185 cohort or cross-sectional studies that recorded SAEs in-vivo during sports participation.

Results: Approximately 5,800 participants were studied in 20 sports using 18 devices that included instrumented helmets, headbands, skin patches, mouthguards and earplugs. Female and youth participants were under-represented and ambiguous results were reported for these populations. The number and magnitude of SAEs were affected by a variety of contributing factors, suggesting sport-specific analyses are needed. For collision sports, being male, being older, and playing in a game (as opposed to a practice), all contributed to being exposed to more SAEs.

Discussion: Several issues were identified across the various sensor technologies, and efforts should focus on harmonising research methods and improving the accuracy of kinematics measurement and impact classification. While the research is more mature for high-school and collegiate male American football players, it is still in its early stages in many other sports and for females and youth populations. The information reported in the summarized work has improved our understanding of the exposure to sport-related head impacts and has enabled the development of prevention strategies, such as rule changes.

Conclusions: Head impact research can help improve our understanding of the acute and chronic effects of head impacts on neurological impairments and brain injury. The field is still growing in many sports, but technological improvements and standardisation of processes are needed.

2.2. Introduction

Concussion, a subset of mild traumatic brain injury, has become a worldwide concern for contact sports participants.[96, 296] A leading reason for this concern is the high incidence rate in various contact sports across various ages and levels of play.[132, 215, 231, 313] While many concussed athletes recover within 14 days following injury,[256] post-concussive symptoms have been reported to persist,[253] and can have negative long-term effects on quality of life.[316] In addition, players with a history of concussions can suffer from long-term impairments such as neurocognitive deficits,[169, 171] memory deficits,[179] behavioural changes,[103, 140] headaches,[103] and altered movement coordination.[40, 259] Concussions in contact sports are frequent, recurrent, and can result in long-term health impairments; therefore, it is important to improve the prevention, identification and management of these injuries.

Concussions are difficult to diagnose using current methods and tools.[256] Such methods rely on the athletes exhibiting detectable signs of a concussion or disclosing their symptoms, but health care professionals face challenges with delayed-onset symptoms[27] and under-reporting by participants.[254] Therefore, objective and reliable tools for identifying concussions or detecting head impacts requiring clinical follow-up are essential.

The mechanics of concussion are generally accepted to be a direct impact to the head, face or neck or an impact elsewhere on the body that transmits a sudden force to the head.[256] The direct impact force or inertial force accelerates the head, leading to brain tissue deformation and strain.[142, 168] Specifically, linear motions (translation of the head) can produce intracranial pressure gradients that can cause focal injuries and angular motions (rotation of the head) can produce shear stresses and tensile strains that can cause diffuse axonal injuries.[202, 321]

As it is currently impractical to measure pressure gradients and brain strains *in-vivo* during sports participation, head impact sensors are used to measure the motion of the head on the assumption that brain strains are in some way proportional to the magnitude of this motion.[149, 281] Sensors have been inserted into helmets,[115] skin patches,[257] or mouthguards[19] and are used in many sports to monitor head impact exposure in games and practices.[115, 155, 266, 378] Head impact exposures have been variously defined as the number of impacts, the magnitude of these impacts and/or the direction/location of the impacts sustained over a defined period such as a game, practice session or full season of sports participation.[26, 91] The number of impacts is a simple count of all head impacts over a certain threshold (often 10g). Impact magnitude has typically been reported as the peak resultant linear and/or angular acceleration[38] or less often as a metric computed from the acceleration signals, such as the Head Injury Criterion (HIC)[402] or the Head Impact Telemetry severity profile (HITsp)[147]. The cumulative burden of head impacts has also been assessed by summing the magnitude (typically peak resultant linear accelerations) of all impacts sustained over a defined time-period.[120] Impact direction and location have received less attention,[310] but may be important because of the different tolerances to brain injuries for different head rotation directions.[143, 385] Because head impact devices have been shown to record "impacts" when in fact no head impact occurred (false positives),[86, 322] the sensor's recordings will hereby be referred to as head acceleration events (HAEs), and the true positives to verified head impacts.

Understanding the number and magnitude of verified head impacts can help better understand concussive injury, identify important risk factors and drive prevention strategies. Thus, head impact sensor data is an important part of helping to achieve the overall goal of reducing impact exposure and the overall risk of concussions. This systematic review was undertaken to summarize the research on *in-vivo* HAEs in sport, to highlight achievements and the current consensus views, and to identify gaps and priorities. It also aims to provide informed recommendations and pathways for future work.

2.3. Methods

This systematic literature review followed the PRISMA guidelines. Records management was performed using a reference management software to import citations, remove duplicates, sort references, and import full texts. PubMed, Web of Science Core Collection, SPORTDiscus, CINAHL and Scopus were searched up to the end of 2019. The methodology utilized keywords in the title, abstract, and keywords fields: ((head injur*) OR (brain injur*) OR concuss*) AND ((head impact*) OR acceler* OR biomechanic*). The search strategy limited results to academic journals, reviews, dissertations, and conference papers.

Articles were included if they met the following two inclusion criteria: (i) the study recorded HAEs *in-vivo* during sports participation, and (ii) the study utilised a cohort or cross-sectional design. There was no restriction of sport, sex, age or level of participation to allow an overall summary to be drawn. Articles were excluded if: (i) the full text was unavailable in English; (ii) no HAE measurement was reported; (iii) they were a review, laboratory study, case report, case series, commentary or opinion piece; or (iv) they were not peer-reviewed. All studies were included, independent of their quality. Duplicate articles were removed and publications were initially screened by the primary author by title and abstract with respect to established inclusion and exclusion criteria. Two additional contributors performed the same search and screening process on the CINAHL database to verify the methods. Additional studies were found in the reference lists in the included publications and relevant reviews, and from the authors' knowledge. The results of the systematic search and screening process are detailed on the PRISMA flowchart in Figure 2.1. The final number of studies included in this review was 185.

All included full-texts were reviewed and relevant data were extracted by the primary author (Table 2.1). A second contributor checked a random sample of 5% of the studies to ensure all relevant data were extracted correctly. Missing information relative to study characteristics was identified. Extracted data are available as electronic supplementary material (electronic table) and published as a spreadsheet with filtering options to allow readers to easily select and visualize the studies of interest. No attempt was made at a meta-analysis of the summary metrics reported by the studies.

A total of 13,453 participants were reported on, however, this included duplicate participants as several studies conducted various analyses on the same cohort. The number of duplicate participants was estimated at 7,618, resulting in 5,835 unique participants. The cohorts are detailed in the supplementary materials (Table A.1 on page 166).

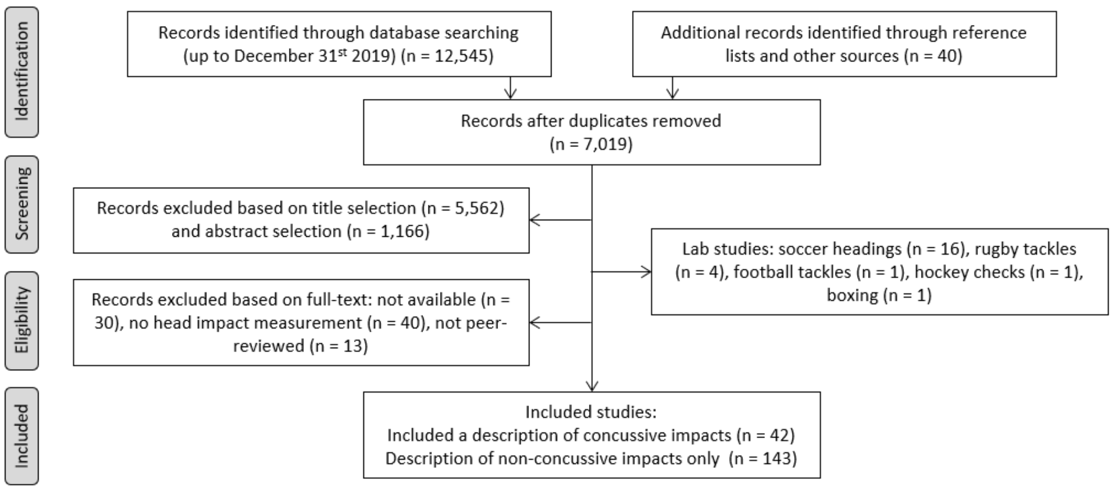


Figure 2.1. Flow of identification, screening, eligibility, and inclusion for the literature review of HAE monitoring.

Table 2.1. List of variables extracted from the included studies and available as supplementary material.

Information	Extraction details
Study characteristics	
Full reference	
Year of publication	
Study design	
Category	'Non-concussive impacts only' or 'Includes a description of concussive impacts'
Methods	
Sport	Sport(s) investigated
Age group	Categorised as <11 years, 11-14 years, High school, Collegiate or >22 years
Males	Number of male participants
Females	Number of female participants
Cohort	Identification of the cohort when reported on in several studies
Technology	Name of the head impact device(s) utilised
Triggering threshold	Threshold used, typically on raw data, to trigger the recording of an impact
Inclusion threshold	Threshold used, typically on processed data, to include impacts for analysis
Video verification	Details on the use of video for verification of HAEs or characterisation of concussive events or impact mechanisms.
Results	
Total number of impacts	Overall number of impacts recorded over the course of the study
Number of concussions	Overall number of concussions observed over the course of the study
Reporting of HAE metrics	Details of how summary metrics were reported (e.g. mean ± SD, 95th percentile, by player position, per game/practice) for linear acceleration, angular acceleration, head velocity, number of impacts, impact location, impact duration, severity metrics, cumulative metrics and other variables.
Risk factors	Summary of risk factors associated with the number, magnitude and location/direction of HAEs, as reported in the included studies with statistical results if available. Includes age, sex, other intrinsic risk factors, event type, player position, impact mechanism and location and other extrinsic factors.
Acute and chronic effects of head impacts exposure	Summary of the analyses on the association between HAE exposure and neurological and physiological performance, e.g. symptoms reporting, cognitive function, white matter integrity.

HAE: head acceleration event, SD: standard deviation.

2.4. Results

Head impact technology

Initial studies monitoring HAEs utilized custom-made equipment[281, 290, 364] until the Head Impact Telemetry (HIT) System (Simbex, Lebanon, NH) was developed. Since the first publication featuring the HIT System in 2005,[115] the number of publications for helmeted sports (i.e., sports requiring the use of hard-shell helmets such as American football, ice hockey or men's lacrosse) increased steadily (Figure 2.2). Head impact studies reporting on non-helmeted sports (no helmet or foam-only headgear, e.g., soccer, rugby, combat sports, women's lacrosse) were fewer with only custom devices being utilized until 2015. Since then, studies on non-helmeted sports have increased as new technologies became more common and accessible, with a maximum of 13 studies in 2017 reporting on six different sports.

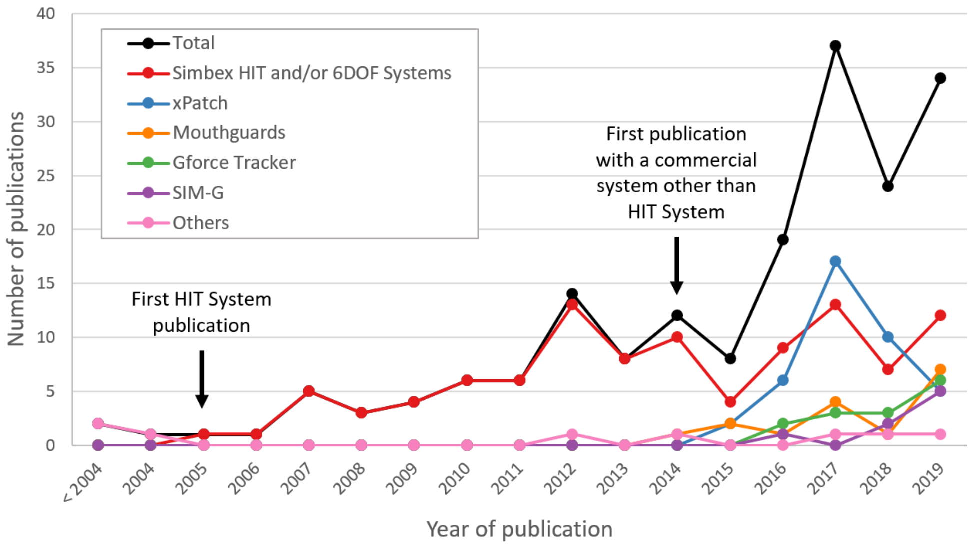


Figure 2.2. Evolution of the number of publications by head impact technology. HIT: Head Impact Telemetry, 6DOF: 6-degree of freedom. Excluding the Simbex HIT and 6DOF Systems, only custom devices (instrumented headgear or earplugs) were reported on before 2014, when the first studies reporting the use of the Intelligent Mouthguard and of the Shockbox were published.

There were 18 different devices used in the included studies; 11 appear to have been commercially available (Table 2.2). Most devices reportedly sampled continuously into a buffer until a specific event (typically linear acceleration exceeding a pre-set threshold) triggered an HAE recording. Several studies specified this triggering threshold and reported this to be set at 5 g (1% of included studies), 7 g (<1%), 9.6 g (3%), 10 g (32%), 14 g (<1%), 14.4 g (12%), 15 g (7%), 16 g (<1%) or 20g (2%) (see Table S1). One study reported recording impacts that generated a linear acceleration over 5 g for 3 ms for soccer headings, which typically yield low magnitude impacts,[339] while the instrumented earplugs[245, 364] were reported to record data continuously. Many studies did not report or did not clearly articulate the method used to trigger impact recordings (39% of included studies).

A secondary threshold was often used on the peak resultant linear acceleration to select HAEs for purposes of analyses. This inclusion threshold varied across studies from 7 g (<1%), 10 g (44%), 15 g (3%), 20 g (6%), 25 g (1%), 30 g (1%), 40 g (4%). Eight studies focused on high-magnitude (> 70 g) or concussive impacts (4%). Fifteen studies (8%) did not use an inclusion threshold and included every recording. The secondary threshold was either not reported or not clearly reported in 51 studies (28%). Both the triggering and the inclusion thresholds were reported in 41% of all studies.

Head kinematics were typically described using peak resultant linear acceleration (PLA) (in 83% of included studies) and peak resultant angular acceleration (PAA) (62%). Some authors opted not to report on angular acceleration measured by the xPatch because of the large measurement errors observed for this device.[412] Other kinematics metrics, such as angular velocity, were reported in less than 10% of studies (see electronic table). The number of HAEs was reported in 79% of studies for various time periods, such as per season or per athlete exposure (AE, any session in which the athlete participated). Impact duration was rarely reported (10% of studies). Several other severity metrics were described in some studies, such as Head Impact Telemetry Severity Profile (HITsp, 27%), Head Injury Criterion (HIC, HIC_{15} or HIC_{36}, 15%) or Gadd Severity Index (GSI, 10%). Cumulative metrics were also utilized (20%) to investigate repetitive HAE exposure and generally consisted of the sum of PLA or PAA over a defined period (e.g. season or AE). The multitude of variables reported, as well as the value reported (e.g., the mean or median), appeared to reflect a lack of consensus on the characteristics of interest.

Table 2.2. Head impact devices utilised in the included studies.

Device name	Studies	Years of use	Technology	Mounting on the head	Sports	References	Independent validation
Head impacts telemetry System (Simbex, Lebanon, NH, USA)	99	2005-2019	Six 1D accelerometers (5DOF, missing the rotation about the vertical axis)	Embedded in helmet liner	American football	5, 10, 13, 23, 26-28, 37, 39, 41, 43, 45-50, 52, 55, 59-64, 78, 81, 90-92, 98-100, 114-116, 120, 127, 134, 135, 147, 149, 151, 153, 157, 167, 181, 187, 188, 193, 194, 229, 244, 248-251, 261, 263, 269, 270, 283, 297, 304, 348-352, 355-358, 360, 373-375, 386, 397-399, 422	IND [93, 180, 369] N-IND [25, 345]
					Ice hockey	34, 114, 152, 248-250, 264-268, 271, 330, 331, 359, 408-410	IND [3, 4, 186]
					Snow sports	110, 111	
				In headbands	Soccer	155	IND [154]
6DOF System (Simbex, Lebanon, NH, USA)	7	2008-2014	Twelve 1D accelerometers (6DOF)	Embedded in helmet liner	Football	97, 116, 346, 350, 351, 422	
				Embedded in boxing headgear	Boxing	378	N-IND [24]
xPatch (X2 Biosystems Inc., Seattle, WA, USA)	40	2015-2019	3D accelerometer and 3D angular rate sensor (6DOF)	Adhered to the skin over the mastoid process	Soccer	15, 76, 77, 236, 257, 284, 292, 322, 328, 334, 335, 381	IND [367]*
					Rugby and Australian football	67, 160, 161, 203, 205, 207, 208, 210, 211, 258, 412	IND [258, 388]
					Lacrosse and hurling	69, 86, 225, 299, 301, 334, 337, 403	IND [363]
					American football	15, 181, 334, 336, 383, 420	IND [93, 396]
					Combat sports	204, 226, 302	
					Cycling	173, 174	
					Ice hockey	118	
Gforce Tracker (GForceTracker Inc., Markham, Ontario, Canada)	14	2016-2019	3D accelerometer and 3D angular rate sensor (6DOF).	Adhered to the inner shell of helmets	Lacrosse	68, 86, 107, 195, 201, 275-277	IND [53]
					American and Canadian football	71-73, 282, 285	IND [58]
				In headbands, on the back of the head	Soccer	158	
SIM-G (Triax Technologies Inc., Norwalk, CT)	8	2016-2019	3D accelerometer and 3D angular rate sensor (6DOF)	In elastic headbands or skull cap and positioned along the nuchal line	Soccer	55, 56, 224	
					Lacrosse	242, 344	
					American football	237, 384	IND [93, 396]
					Water polo	70	
Vector mouthguard (Athlete Intelligence, Kirkland, WA, USA)	7	2016-2019	3D accelerometer and 3D angular rate sensor (6DOF)	Boil-and-bite fit	American football	129, 170, 190-192, 353, 424	

6DOF: six degree-of-freedom; 1D: one dimension; 3D: three dimensions, IND: independent validation study, N-IND: non-independent validation studies from manufacturers.

*: this study was conducted on post-mortem human subjects.

Table 2.2 (continued)

Device name	Studies	Years of use	Technology	Mounting on the head	Sports	References	Independent validation
Stanford mouthguard	5	2015-2019	3D accelerometer and 3D angular rate sensor (6DOF). Different designs and sensors over the years.	Custom fit	American football Combat sports Soccer Rugby union	[164, 220, 416] [164, 300] [273] [300]	
Intelligent Mouthguard (Prevent Biometrics, Edina, MN, USA)	2	2014, 2019	Two sensing packages: 3D accelerometer and 3D angular rate sensor (6DOF), or four 3D accelerometers (6DOF)	Custom fit or boil-and-bite	American football, Boxing	[19, 20]	N-IND [18-20, 162]
xGuard (or DVT3, X2Biosystems Inc., Seattle, WA, USA)	1	2015	3D accelerometer and 3D angular rate sensor (6DOF)	Custom fit	Rugby union	[209]	IND [57, 369]
Custom mouthpiece	2	2019	3D accelerometer and 3D angular rate sensor (6DOF). Different versions. Sensor and battery component of an xPatch device.	Custom fit	Soccer	[273, 339]	N-IND [272, 339]
Custom earplugs	1	2004	3D accelerometer (3DOF, linear only)	One ear only	Boxing	[364]	
Custom earplugs	1	2012	Three 1D accelerometers and 3 1D angular rate sensors (6DOF)	Accelerometers in one ear, angular rate sensors in the other	Rodeo	[245]	
Custom headgear	1	2000	3D accelerometer (3DOF, linear only)	Sensor implanted in the padding of the helmet	Ice hockey, American football, soccer headings	[290]	
Custom headband	1	1971	Three 1D accelerometers (3DOF, linear only)	Headband	American football	[281]	
BodiTrak (HeadHealth Network)	1	2017	Estimation of linear acceleration from pressure sensors	Helmet-based sensors	American football	[372]	
Reebok Checklight™	1	2018	3D accelerometer and 3D angular rate sensor (6DOF). Light system to indicate impact magnitude.	Skull cap	American football	[156]	IND [93]
Shockbox (Impakt Protective Inc., Canada)	1	2014	Non-accelerometer based impact sensor	Adhered to the outer surface of a helmet	American football	[414]	IND [93]
Skin patch (not reported)	1	2019	Not reported	Not reported	Combat training	[343]	

6DOF: six degree-of-freedom; 1D: one dimension; 3D: three dimensions, IND: independent validation study, N-IND: non-independent validation studies from manufacturers.
*: this study was conducted on post-mortem human subjects.

Demographics of participants

Of all 185 reviewed studies, more studies reported on helmeted than non-helmeted sports (75% vs. 25%). The most commonly reported sports were American football (57% of the included studies), soccer (10%) and ice hockey (9%).

The following results reference the 5,835 unique participants, after excluding duplicates across studies. The number of participants per study (excluding the compilation studies) ranged from three[204] to 185,[269] with a mean of 37 and a median of 31 [interquartile range 17 – 47]. The largest groups studied were male American football players (total of 3,650 unique players) and male lacrosse players (N = 397) from the 5,627 unique participants for whom both sport and sex were reported (Figure 2.3). Of the 185 studies reviewed, 22% included one or more female participants (range 1-58). Females represented less than 15% of the overall investigated population (Figure 2.3, Figure 2.4). They were best represented in soccer (N = 366), ice hockey (N = 172), and lacrosse (N = 99).

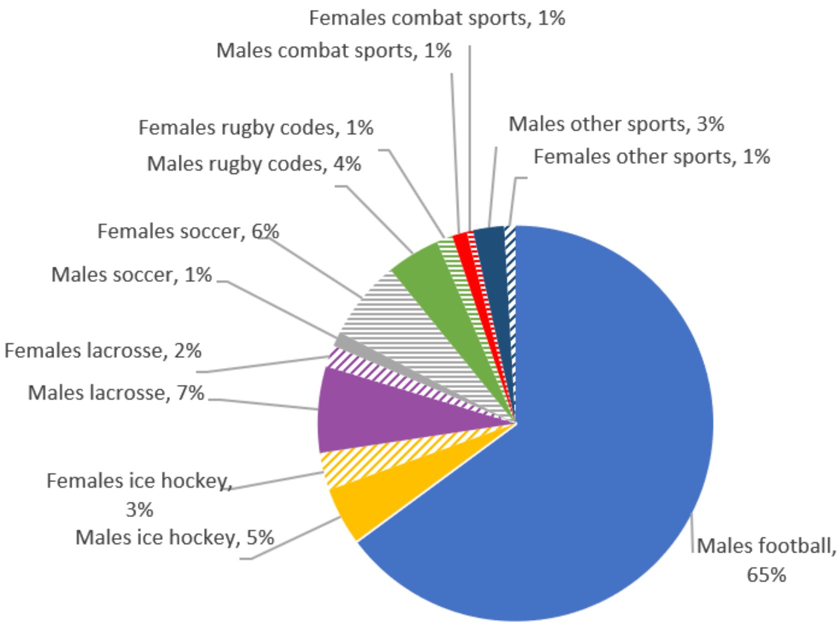


Figure 2.3. Percentages of male and female participants by sports category in the included studies of HAE monitoring. Studies were grouped by type of sport. The football category included 107 American and three Canadian football studies. The Lacrosse category included 16 Lacrosse and one hurling studies. The combat sports category included five boxing, two mixed martial arts, one taekwondo, one sparring and one wrestling studies. The rugby codes category included five Australian Rules football, five rugby league, three rugby union, and one flag football studies. Other sports included two snow sports, one downhill mountain biking, one BMX, one rodeo, and one water polo studies. Percentages were calculated based on 5,627 unique participants.

Collegiate (18–22 years) and high school (14–18 years) age groups contributed 77% of the overall studied population, and only 16% and 7% of the participants were from the youth (<14 years) and adult (>22 years) age groups (Figure 2.4). There were 206 participants younger than 11 years reported in eight studies.[59, 60, 63, 97, 174, 207, 210, 237, 398, 420, 422]

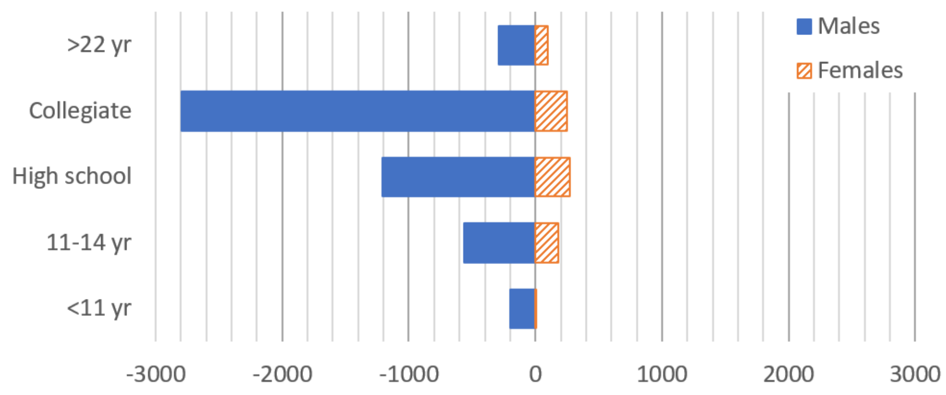


Figure 2.4. Age and sex distribution across the individual participants from the studies that reported sufficient sex and age details. Age groups were often reported as educational level. When the mean age was reported rather than the educational level, the studies were grouped under collegiate (mean age >18 and ≤22 years) or high school (>14 and ≤18 years)

Risk factors

Intrinsic Risk Factors for Head Impacts

Twelve of the included studies[34, 77, 118, 205, 225, 271, 335, 337, 408, 410, 412] directly investigated sex differences using the same sensor and methods for both sexes within a single study (Table 2.3a). Female athletes sustained fewer HAEs than male athletes in sports where body checking or tackling is allowed. Sex differences in the magnitude of linear acceleration were equivocal for all sports. Females sustained lower angular magnitude HAEs than males in ice hockey, but there were insufficient data for the other main sports. Lower numbers of HAEs and lower magnitudes were also reported for women in boxing.[378]

Eleven studies[60, 77, 161, 174, 194, 203, 257, 330, 360, 398, 420] compared age groups using the same sensors and methods for all age groups within a single study (Table 2.3b, Table 2.3c). Most studies reported that younger players experienced fewer and lower magnitude HAEs. Two exceptions were Australian rules football, where more high-magnitude impacts were observed for high school athletes than collegiate athletes, and BMX, where the number and magnitude of accelerations >5 g decreased with increased age (from 6 to 18 years)[174]

Two studies found a positive association between body mass index (BMI) and both the number and the magnitude of HAEs,[330, 420] while a third study did not find significant differences[331] (Table 2.3d). The results from the studies evaluating neck strength did not suggest strong associations with the magnitude of HAEs for high school and collegiate athletes[195, 268, 355] (Table 2.3e). Other intrinsic risk factors were investigated with respect to the number or magnitude of HAEs in one or two studies, including neck neuromuscular properties, visual performance, playing experience, maturity offset, safe-play knowledge and behavioural patterns, movement competency, body temperature, preparedness, and ethnicity (electronic table, "Other intrinsic factors"). No strong, consistent correlation was found for any of these other intrinsic factors.

Extrinsic Risk Factors for Head Impacts

For collision sports (e.g., American football and ice hockey), games generated more and higher magnitude HAEs than practices, reflecting the intensity of game play[47] (Table 2.3f). For lacrosse and soccer, more HAEs occurred in games than in practices for male athletes, but the trend was for equal magnitude and lower numbers for female athletes. Therefore, special attention may be warranted for practice sessions in these sports and populations.

Numerous studies of American football compared the number and magnitude of HAEs by event type (game, scrimmage, full-contact practice, limited-contact practice), type of play (offensive, defensive, special play), or drill at the collegiate,[10, 263, 304, 336, 373, 383] high school,[45, 49, 244] and youth[63, 193, 237, 398] levels of participation. Findings from these studies have led to, or have the potential to guide various positive changes, such as the regulation of the number of weekly full-contact practices or specific drills (e.g., the King of the Circle drills) (electronic table, "Event type").

Differences in the number and magnitude of head impacts across player positions were investigated in American football,[39, 45, 47, 49, 60, 64, 90-92, 116, 135, 153, 244, 263, 270, 355, 358, 360, 373] ice hockey,[118, 266, 267, 330, 331, 408] lacrosse,[69, 201, 242] rugby codes,[208, 209, 211] soccer,[224, 236, 322] Australian Rules Football,[161, 203] Canadian football,[282] and water polo[70]. Most of the studies of American football, rugby codes and Australian rules football found differences across player positions, whereas the findings were equivocal for ice hockey, soccer and lacrosse.

Differences in the number and magnitude of HAEs at different stages of a game were observed in American football,[357] soccer,[236] and rugby.[211] The second half of the games were associated with a greater number of impacts[211] and a greater proportion of impacts with high PLAs[211, 236] and PAAs.[211] Some researchers have postulated that player fatigue is responsible for these intra-game differences, although other factors (e.g., end-of-game pressure to win) may play a role and warrants further investigation.

Contact mechanics (e.g. player-to-player, head-to-ground, or heading the ball) affected HAE magnitude in American football,[5, 61, 304, 357, 398] ice hockey,[264, 265, 410] lacrosse,[68, 201, 225, 403] soccer,[56, 158, 224, 273, 292] rugby,[67] and Australian rules football.[258] A better understanding of impact mechanics can motivate rules changes, such as the regulations related to body checking or intentional head-to-head contact in some collision sports.

Other extrinsic factors were investigated by individual studies (electronic table, "Other extrinsic factors"). These factors included team variability, the wearing of a jugular vein compression collar, foul play, ambient temperature, helmet and faceguard types, ball possession, number of two-a-day practices, tackling technique, level of participation, and time of the season. Many of these factors would benefit from further research, or from research being extended to other sports.

Comparison of studies

The feasibility of a detailed meta-analysis of the included studies was limited by the different sports, sensors, methodologies, and type of data reported. For male American football players wearing the HITS sensors, the data seemed to confirm that impact magnitude increased with age in youth players[59, 63, 81, 98, 397, 422] although no statistical analysis was conducted (Figure A.1 on page 167). Somewhat surprisingly, the data also suggested that youth players sustain higher magnitude impacts than collegiate players.[127, 194, 263, 398, 399] Similar comparisons undertaken for other sports did not add any generalizable information as few studies were quantitatively comparable.

Table 2.3. Results of the sex, age, BMI, neck strength and event type comparisons of number and magnitude of HAEs by sport. A higher or lower result is representative of a significant difference as per the study's methods; a similar result means the difference was tested by the authors and resulted in a non-significant difference.

	American and Canadian Football	Ice hockey	Lacrosse	Rugby and Australian rules football	Soccer
a) Sex (for females, with males as the reference)					
Number of HAEs	-	-	Similar [337]	-	Similar [77, 335]
	-	Lower [34, 118, 271, 408]	Lower [225]	Lower [205, 208, 211, 412]	-
Magnitude of linear accelerations	-	Higher [118, 271]	Higher [225]	-	Higher [77]
	-	Similar [410]	-	Similar [412]	Similar [335]
	-	Lower [34, 408]	Lower [337]	Lower [205]	-
Magnitude of angular accelerations	-	-	-	Higher [205]	-
	-	Similar [271]	-	-	Similar [335]
	-	Lower [34, 118, 408, 410]	Lower [337]	-	-
Other metrics	-	Lower HITsp [408, 410]	-	-	-
	-	Similar HITsp [118]	-	-	-
b) Age (for high school athletes, with collegiate athletes as the reference)					
Number of HAEs	-	-	-	Similar [161]	-
	Lower low-magnitude impacts [360]	-	-	-	Lower [257]
Magnitude of linear accelerations	-	-	-	-	Lower [257]
Other metrics	-	-	-	More high-HITsp impacts [161]	Lower cumulative PLA and PAA [257]
c) Age (youth players - for younger players, with older players as the reference)					
Number of HAEs	Less high-magnitude impacts [60]	-	-	-	Lower [77]
Magnitude of linear accelerations	Similar [398]	-	-	-	-
	Lower [194, 420]	-	-	-	-
Magnitude of angular accelerations	Similar [194, 398]	-	-	-	-
	Lower [420]	Lower [330]	-	-	-
Other metrics	-	Lower HITsp [330]	-	-	-

PLA: peak linear acceleration; PAA: peak angular acceleration; HITsp: Head impact telemetry severity profile; HIC: head impact criteria.

Table 2.3. (continued)

	American and Canadian Football	Ice hockey	Lacrosse	Rugby and Australian rules football	Soccer
d) BMI (higher BMI, with lower BMI as the reference)					
Number of HAEs	-	Higher [330]	-	-	-
	-	Similar [331]	-	-	-
Magnitude of linear accelerations	Higher [420]	Similar [331]	-	-	-
Magnitude of angular accelerations	Higher [420]	Similar [331]	-	-	-
e) Neck strength (weaker-neck athletes, when compared with stronger-neck athletes)					
Magnitude of linear accelerations	-	-	Similar [195]	-	-
Magnitude of angular accelerations	-	-	Similar [195]	-	-
Other metrics	Similar odds of sustaining mild, moderate and severe impacts [355]	Lower HITsp [268]	-	-	-
f) Event type (in games, with practices as the reference)					
Number of HAEs	Higher [45, 47, 60, 90, 98, 194, 237, 282, 373, 397, 414]	Higher [118, 359, 408]	Higher [337]	-	Higher [335]
		Higher high-magnitude impacts [266]	Higher for men [225]		Higher high-magnitude impacts for males [335]
	Similar [81, 422]	-	Similar [242]	-	-
	-	-	Lower for women [225]	-	Lower high-magnitude impacts for females [236]
Magnitude of linear accelerations	Higher [45, 47, 81, 129, 244, 282]	Higher [266]	-	-	-
	Similar [98]	Similar [267]	Similar [242, 337, 344]	-	Similar [224, 335]
	Lower [263, 422]		-	-	-
Magnitude of angular accelerations	Higher [45, 47, 81, 129, 244]	Higher [267]	-	-	-
	Similar [98]		Similar [242, 337, 344]	-	Similar [224, 335]
Other metrics	Higher HITsp [45, 92]	Higher HITsp [267]	Higher cumulative PLA and PAA [337]	-	-
	Higher HIC15 [129]	-	-	-	-

PLA: peak linear acceleration; PAA: peak angular acceleration; HITsp: Head impact telemetry severity profile; HIC: head impact criteria.

2.5. Discussion

This systematic review identified 185 studies and highlighted that the number of publications and devices available grew rapidly over the last 15 years, reflecting an increasing need to understand head impacts and develop prevention strategies. As a result of this rapid development, there were large disparities in methodologies for data collection and reporting that would benefit from standardisation. Nonetheless, sport-specific research has substantially improved our understanding of sport-related head impacts, and in some cases has driven changes aimed to protect the athletes. However, female and youth populations across all sports are under-represented and their exposure remains insufficiently understood.

Head impact technology capabilities

Most commercially-available sensors have undergone independent validation (Table 2.2). Under controlled settings in the laboratory, measurement errors were observed when compared with reference sensors and have been summarized elsewhere.[307] These errors likely increase when these sensors are used by athletes on the field in a real game or practice situation. Specifically, the sensor's fit on the head and mechanical coupling with the skull,[4, 180, 367, 418] as well as variability in head morphology[273] may increase the errors. Assessing the accuracy of the sensors *in-vivo* is particularly difficult because of the lack of a reference sensor but is of high importance in the context of understanding and potentially identifying individual injuries.

Several different triggering and inclusion thresholds have been used, and these thresholds affect the number of impacts recorded and the summary metrics reported.[206, 220] For example, using a 14.4 g triggering threshold rather than a 10 g threshold has been shown to result in 42% of impacts not being recorded and an increase of up to 50% in summary metrics (e.g., a cohort's median PLA).[206] Any variable that uses the number of impacts recorded – including mean and median PLA or PAA values – or any sort of cumulative value is directly affected by the triggering and inclusion thresholds. It is therefore not advisable to compare impact numbers, magnitude averages/medians or cumulative measures between studies that used different thresholds.

The choice of triggering thresholds is complex as it balances the technology's capabilities with user needs. It is often chosen to optimize data capture,[339] minimise data storage,[164] maximise battery life[164] or reduce the number of HAEs to focus on higher magnitude HAEs.[351] It has been suggested that a higher threshold (30-34 g) would optimise the capacity of the device's algorithms to correctly identify actual head impacts and ignore spurious recordings.[322, 412] However, there are concerns that this value is too high to capture all the HAEs that matter in the assessment of an athlete's cumulative exposure. This range of values is also above the lowest peak resultant linear accelerations observed for a concussion-inducing impact (25.2 g).[374] A recent soccer study used a lower threshold (5 g) but required that the signal remains about the threshold for more than 3 ms to capture low-magnitude events that may be of interest.[339] Because the accumulated effects of low-magnitude impacts on the brain and on the occurrence of concussion are still not fully understood,[47] it is important to continue recording low severity impacts. This may be difficult to achieve in sports with a high number of HAEs when devices have limited battery life and storage capacity. Therefore, threshold choice may need to be sport-specific until the technology allows the recording of more HAEs.

In many instances, assessing the number of impacts sustained, independent of the magnitude, is of interest. To facilitate this assessment, sensor systems typically incorporate proprietary algorithms that classify each recorded HAE into valid impact (i.e., a head, neck, or body impact) or invalid event (e.g., artefact due to manipulation of the device). Differentiating between valid and invalid acceleration events is essential as they present different characteristics, particularly in signal frequency content.[19, 273, 343, 416] The algorithms are not typically known of the end-user[258] and their performance has been questioned. Video analysis has shown that between 20 and 89% of all recorded HAEs classified as valid by the devices proved to be artefacts (false positive)[67, 86, 220, 224, 258, 339, 403] and between 14 and 40% of valid impacts were improperly discarded (false negative).[258, 292, 322, 416] In addition, 26 to 31% of impacts visible on video were missed by the sensors.[67, 220, 258, 339] Because of these detection and classification issues, the quality of many published datasets, especially those not verified by video (84% of the included studies did not systematically verify HAEs), remains uncertain and their summary metrics may be tarnished.[292]

To improve event classification, some mouthguards have introduced in-mouth sensing to automatically remove HAEs that were recorded while the mouthguard was not positioned against the teeth. Laboratory results showed an improvement in classification accuracy from 65% to 94% when using in-mouth sensing.[419] In-mouth sensing was reportedly available on the Stanford[220, 416] and Vector mouthguards.[170, 424] Machine learning algorithms trained on video-verified[416] or visually-verified head impacts[343] have resulted in better accuracy when compared with the manufacturer's algorithm (from 61% to 88%),[343] up to 94%.[416] Until these solutions have undergone more testing on other devices and been further validated, it is suggested that researchers use video footage to systematically verify impacts.

Video analysis not only allows the identification of true positives, but it can also be used to verify impact location and direction and to analyse impact mechanics. However, several challenges and limitations have been reported by some authors. Firstly, there were reports of issues with HAE data and video synchronisation[258] and researchers resorted to time calibration steps at the start and/or end of each session.[68, 69, 86, 195, 220, 292, 410] Secondly, several good quality cameras are needed to have a full view of the playing field to optimise verification of impacts.[220, 322, 412] Even with multiple cameras, some impacts will be obstructed by other players,[209] and some collision events will trigger several HAEs within a short period and therefore may not be distinguishable in the video.[220] Finally, video verification, especially if looking for head and body impacts on video in addition to matching HAEs, is a highly time-consuming task. For these reasons, efforts should be put into developing accurate impact classification algorithms.

Most commercially-available head impact devices do not provide the user with raw signals, leaving researchers with few options for data processing and analysis. There is limited information about how the sensor signals are processed (e.g., filtering, differentiating, inclusion threshold, etc.) and how differences in processing methods affect the end results.[258] The head impact sensor manufacturers could disclose their processing methods and algorithms to allow researchers to better interpret the data and facilitate comparison with other studies and sensors. Alternatively, sensor manufacturers could provide a way for researchers to access the raw sensor signals.

Future research would benefit from standardised data collection methods and better reporting of the devices' uses and setup. To facilitate this, two sets of Common Data Elements are available from the National Institute of Neurological Disorders and Stroke database. These cover head kinematics estimates[287] and video device confirmation,[288] both citing the works of Cortes et al.[86] and Kuo et al.[220] Missing from these lists is the inclusion threshold, when used in addition to a triggering threshold. Standardised reporting of the summary metrics as median and interquartile ranges is also recommended because the data is not normally distributed.[206]

Epidemiological gaps

Variety of Sports

The amount of research conducted on American football was substantially greater than that of other sports. This disproportionate attention may have been influenced by increased media awareness of neurological disorders secondary to professional football. Technology may also have played a role in favouring the study of helmeted sports as the systems were initially limited to helmet-embedded systems. Other sports, particularly rugby and ice hockey, present concussion injury rates higher than or similar to American football;[313, 323] there are also many times more soccer and rugby players worldwide than American football players.[112, 125, 415] With the technology to explore non-helmeted sports now available, more research is warranted on those sports that present high concussion risks and higher participation numbers.

Sex of the Participants

It was clear that in collision sports (ice hockey, rugby, Australian football and boxing), males experienced more and higher magnitude HAEs than females. However, sex-differences were not as well studied in other sports where most HAEs are from incidental collisions or heading the ball, and where female athletes may be more represented. Globally, the poor representation of females in HAE studies did not reflect the latest epidemiological results reporting increased concussion risks,[323, 333] higher symptoms severity[333] and longer recovery periods[138] for women participating in sports, when compared with men. The under-representation of females in HAE monitoring studies may be attributed to the low proportion of women participating in helmeted sports. However, women's participation in soccer, basketball, or rugby is high and on the rise, and there are, for example, six times as many female soccer players as male American football players worldwide.[112, 125] A better assessment of the risks for female athletes, especially in sports where rules are the same for both sexes,[205] is needed to understand why females may be more sensitive than males to the negative effects of HAEs.

Age-related Differences

Youth and adolescent participants are also of concern for concussion for several reasons. Evidence in American football showed that because of higher numbers of participants in those age groups, about 99,000 youths and 79,640 high school players would sustain a concussion each year, versus only 3,905 collegiate players.[112] Adolescents and youth players also go through longer recovery periods when compared with adults[150, 260] as the pathophysiological response of the developing brain is more detrimental than that of the more mature adult brain.[22]

Differences in head kinematics observed in the included studies between adults and children are thought to be influenced by morphological differences. A higher head-to-body mass ratio and a weaker neck[119] suggests that younger players would have higher head kinematics. This was verified in BMX where the decreasing number of HAEs with increasing age was supposedly associated with the development of neck and shoulder musculature.[174] However, other studies included in this review showed the opposite trend, wherein both the number and magnitude of HAEs increased with age. These contradictory observations may be explained by the interactions of several intrinsic (age, morphology, BMI, strength level and playing experience) and extrinsic factors (rules, coaching style, level and intensity of play).[81, 194] As a result, it is difficult to isolate the effects of age, as younger players are generally smaller, lighter, less experienced, and would play at lighter intensity levels than older players. Strategies are in place in some sports to group young athletes not only by age, but also by weight, and for the sports where this is not the case, this type of separation is worthy of investigation. Additionally, neck and upper back strength in youth athletes is an area of research that has not been explored yet and warrants future efforts.

Uses of HAEs research

While the possibilities for comparison between studies are limited for methodological reasons,[124] the measurement of head kinematics has led to several outcomes. Four levels of outcomes were identified from the included studies and other related studies.

Level 1 – Single impacts

Finding an association between single HAEs and diagnosed concussions was one of the objectives in the development of head impact sensors. Attempts have been made to determine concussion risk thresholds based on PLA and PAA,[134, 135, 148, 351, 423] even though concussions have occurred across a wide range of acceleration magnitudes, and without regard to impact direction.[38, 149] The sensors' validity for PLA measurements has been better than the validity for PAA measurements[367, 369] even though diffuse brain injuries, including concussion, are thought to be more closely tied to angular head kinematics than linear head kinematics.[182, 212] Other measures of head kinematics (e.g., head angular velocity change) may be equally well correlated to brain injury and are more directly (and perhaps more reliably) measured by the current sensing technologies.

Head impact monitoring systems, in conjunction with proposed thresholds, have been reported to show a high proportion of false positives (i.e., an HAE above a specified injury threshold that did not result in a concussion).[269] If head impact sensors in their current state were to be used to send alerts to potentially concussed players based on the number and/or magnitude of HAE measures, athletes may become desensitized to the repeated false alerts and, conversely, real injuries could go unnoticed.[269] The lack of a strong relationship between HAE and concussion[269, 352] might be related to issues such as the diagnosis process,[44, 326] sports participants underreporting symptoms,[233, 254] the effects of intrinsic and extrinsic risk factors,[2, 122] and sensor validity. Based on current state-of-the-art technologies, athletes and team medical personnel should not at this time rely on head impact devices alone to detect concussion. The signs and symptoms of concussion (either noticed by spotters, referees, coaches; or reported by the players) should continue to remain central to physician assessments.

On the other hand, while there is no evidence that the devices alone provide data that could independently and reliably identify concussed athletes,[256, 269] the HAE data has strong potential for increasing our understanding of injury mechanics. There are, for example, several questions being investigated using HAE data as input parameters for numerical models that quantify brain tissue responses to various real impact conditions. These models could allow the study of the effects of impact location and direction, head morphology, type of impact (for instance helmet-to-helmet vs. helmet-to-stick), or protective equipment on the development of brain damage.[28, [87, 164, 250, 274] However, at this stage and for this type of analysis, the level of error for single impacts measured by the devices is of concern.

Level 2 – Exposure from a single session

Related to the diagnosis of concussion, there is a distinction between a concussive impact, where a player sustains one particularly damaging impact, and a concussed athlete, when there is no single standout impact but rather the accumulation of several more moderate events over a single session.[26, 27, 92, 114] This phenomenon is illustrated by a high proportion of concussed athletes presenting delayed-onset symptoms, in opposition to athletes who were immediately removed from play (57% vs. 43%).[27] The effects of repeated HAEs over a single session have been investigated (electronic table, "Acute and chronic effects of head impacts exposure"). Most studies measured significant changes in symptoms, blood biomarkers or cognitive performance between pre- and post-session assessments. In several cases, there was also a positive association between these changes and the number and/or cumulative magnitude of HAEs sustained, highlighting the deleterious effects of repeated head impacts. Reviews have been published to summarize some of these associations and provide guidance for future research.[172, 361] This branch of research is essential to understand how to monitor athlete exposures, and possibly promote changes to the organisation of sporting sessions, such as the duration of games or substitutes turn-over, or the time needed between sessions. It is important that sensors accurately report the number and severity of valid head impacts recorded over the short periods of time that a practice session or a game represent.

Level 3 – Group-wide analyses

Despite the limitations identified with the devices concerning the number and magnitude of HAEs, the data collected over the years has substantially increased our understanding of the demands of the sports and has driven changes in playing rules and organisation of practices. American football has seen most of these changes, including rules decreasing the authorized closing distance (which affects a player's speed at impact)[304, 357, 398] and the number of players involved in a tackle,[357] and regulations limiting the weekly number of full-contact sessions,[49, 373] the types of drills[10, 63, 193] or the use of protective equipment during practices.[263, 336, 383] Ice hockey has also been influenced by regulating body checking for youth players.[264, 265] For sports where head impact research started later, similar changes are expected as new research identifies risky behaviour and assesses prevention strategies.

From the perspective of group-wide analyses, the inaccuracy in a sensor's measurements may be acceptable if that inaccuracy is distributed uniformly across the dataset. This phenomenon, however, is difficult to evaluate because many factors can influence the errors.[307] Studies must also account other sources of bias, such as variability between teams[81, 90, 92] that can be associated with coaching and/or playing style.[244, 271, 398] Therefore, rule changes should be based on, and analysed, using data acquired from homogeneous pools of athletes wearing the same sensors.

Level 4 – Lifetime exposure

Our review has focused on cohorts or cross-sectional studies that directly measured head impacts during sports participation. The data collected for these studies have formed an important basis for potential metrics on the cumulative burden of long-term exposure. Cumulative burden metrics have been developed, such as the cumulative head impact index (CHII).[280] The CHII estimates lifetime head-impact exposure for American football players and has been associated with later-life cognitive, neurologic and behavioural impairments[280] as well as fluid biomarkers levels.[7, 8]

Limitations of the review

This review has several limitations, the main one being the large number of disparate studies included. Given the amount and complexity of the information presented in these studies, the current summary was confined to a relatively high level. For instance, differences in the number and magnitude of HAEs were observed between playing positions or playing groups in different sports; sport-specific analyses are needed to address these differences. Also, while differences observed in the various studies were reported (Table 2.3), no re-analysis of the summary metrics pooled across multiple studies was conducted. Re-analysis of pooled data might shed light on some of the small between-group differences reported in some studies, e.g. a difference in mean PLA of less than 2 g between full-contact practices and games.[263] Given the measurement errors reported in some validation studies, these differences, although statistically significant, may not be practically or clinically relevant.

2.6. Conclusion

The development of head impact technology has enabled the measurement of millions of HAEs in multiple sports, providing information about their number, magnitude and direction. This information has substantially increased our understanding of the risks athletes are exposed to during sports participation, on four different levels: (1) single HAEs have been used to study the mechanisms of concussive injury and the brain tissue response; (2) the number and magnitude of repeated HAEs over a single session have been shown to have acute deleterious effects on neurological performance, which could be mitigated by rule changes, playing time, or protective equipment; (3) season-long assessments have allowed the burdens associated with type of play, impact mechanisms, or playing position to be quantified and have enabled changes in rules and practice management in some sports; and (4) the lifetime exposure has been estimated and utilised to better understand later-life health impairments.

A large proportion of the research, and thus improvements in managing head impacts, has occurred in American football. Other sports, such as soccer or rugby, can benefit from the knowledge gained by this research but more sport-specific research is needed. Female and youth athletes were underrepresented despite their large participation numbers and their accrued vulnerability to head injury. Depending on the sport, researchers may want to focus on different aspects, such as understanding impact mechanics and exposure during practice, or neck neuromuscular development for these weaker groups.

While HAE research is useful on many levels, there are several caveats associated with the use of the technology *in-vivo*. In particular, because head impacts have been shown to present signal characteristics different than acceleration events artefacts, poor discrimination of valid from invalid HAEs pollutes the dataset in ways that have not been fully evaluated. Efforts should focus on improving the classification algorithms to increase trust in the dataset and limit cost- and time-intensive data verification steps. Finally, standardized ways to measure, process, analyse and report HAE data would benefit the field and allow robust meta-analyses to be performed.

Chapter 3 - Laboratory evaluation of the CSx instrumented mouthguard for measuring head impact biomechanics

This chapter comprises the manuscript submitted for publication:

> Le Flao, E., Verma, K., Bourdet, N., Siegmund, G. P., Willinger, R., & Borotkanics, R. J. (2020). Laboratory validity of the CSx instrumented mouthguard for measuring head impact biomechanics. *(Submitted February 2021)*.

Supplementary materials are presented in Appendix B.

3.1. Prelude

The previous chapter highlighted that wearable technology has been developed and used to measure head impacts during sports participation. However, head impact sensors have been shown to lack measurement accuracy, leading some authors to discard data collected on the field. Therefore, sensors should ideally be independently assessed before being used *in-vivo* to ensure that their accuracy to measure kinematic data is sufficient for the intended use. In Chapter 3, we evaluate the laboratory validity and accuracy of the CSx mouthguard, compared with reference sensors. While previous studies have focused on the peak resultant accelerations, this chapter went beyond the traditional discrete metrics and used the CORA method to compare reference and experimental time series.

3.2. Abstract

Background: Accurate sensors are needed to assess the number and magnitude of *in-vivo* sports-related head impacts.

Objectives: The aim of this study was to evaluate the CSx instrumented mouthguards over a range of impact conditions.

Methods: First, uniaxial linear and angular drop tests were conducted. Second, the mouthguards were affixed inside an instrumented headform subjected to free drops. Comparative analyses were performed on the peak and area under the curve. Validity was achieved if a robust regression model's slope and intercept were not significantly different from 1 and 0, respectively, and the mean absolute error (MAE) was below 10%. The CORA analytics was used to assess the time series.

Results: Peak linear acceleration showed limited validity with most impact directions leading to intercepts and slopes significantly different from 1 and 0, respectively, and MAE averaged 13 ±8%

(range 0-38%). Peak angular velocity behaved better on the linear regressions but still showed limited accuracy (MAE: 14 ±8%, 1-57%). There was strong agreement in terms of signal shape (CORA cross-correlation ratings ≥ 0.91) but large differences in size (CORA size ratings 0.41-0.88) between CSx and the reference.

Discussion: Overall, the linear acceleration peaks and AUCs were not estimated with enough accuracy to satisfy our definition of validity. A temporal misalignment between the CSx linear and angular data contributed to moderate overall CORA scores for the angular data. Testing a larger number of mouthguards samples and investigating the accuracy of the signals along the three axes of measurement rather than the resultant signal may help further understand the source of errors.

Conclusions: The results indicated that the CSx mouthguards performed similarly or better than other instrumented mouthguards, but that the analysis of individual impacts peaks should be done with caution nonetheless.

3.3. Introduction

Head impacts in sports generate head accelerations that deform the brain and have the potential to cause concussions.[256] It is possible to measure the number and magnitude of these impacts *in-vivo* with head impact sensors installed in helmets,[115] mouthguards,[209] headbands,[56] ear plugs[364] or skin patches.[257] Such devices typically integrate accelerometers and angular rate sensors and record short bursts (40-100 ms) of data every time a threshold is crossed, e.g., when one axis of linear acceleration exceeds 10 g. Like concussions, this typically occur when an athlete is hit in the head directly or when an inertial force is transmitted to the head from a hit elsewhere on their body.

Head impact sensors were developed in part to provide sideline personnel and health care professionals with tangible exposure data that could be related to brain damage. Because of their relatively simple design and low cost, head impact sensors have become readily accessible and have been deployed in many sports before the relationship between exposure and concussion could be strongly established.[269] Therefore, their clinical utility to health care professionals in identifying concussions remains a subject of debate.[256, 269]

To be of use in identifying a potentially concussed athlete, head impact devices must be valid - it must measure what a reference system measures with accuracy, i.e., close to the true or accepted value. Many studies have endeavoured to assess these devices' validity in the laboratory and have given cause for concern. For example, while showing good overall correlations with reference measurements,[25, 345] the Head Impact Telemetry (HIT) system, the first and most popular helmet-based device, demonstrated errors in linear acceleration (LA) and angular acceleration (AA) of up to 60% at some impact locations.[3, 369] These errors are thought to be caused by relative movement of the helmet on the headform.[184] Skin-mounted sensors are also affected by poor coupling due to soft tissue movement. In a study comparing skin patches to mouthguard-based devices, the measurement errors reached 120% for linear acceleration and 290% for angular acceleration.[418] Furthermore, when compared to reference sensors rigidly fastened to cadaver heads, the linear and angular accelerations were overestimated by the skin patches by 64% and 370%, respectively.[367] Because mouthguards are fitted to the upper dentition, they are more rigidly coupled to the skull than helmet- and skin-based devices and they have been suggested to be generally more valid.[165, 418]

The limited validity[298, 369] of head impact sensors is a challenge, particularly at a time where there is a strong demand for objective tools to help assess and potentially identify concussion,[269] and a wide range of available wearable sensors. For every new sensor being developed and used to measure *in-vivo* head impacts, the assessment of its capacity to accurately quantify skull kinematics is needed. Measurement errors have also been shown to be affected by impact magnitude[345] and impact location.[369] Consequently, it is important to assess a device's validity for a range of impact conditions that are realistic with respect to how this device is to be used during sports participation. Accordingly, the aim of this study was to evaluate a new instrumented mouthguard in a laboratory setting, over a set of defined experimental conditions. To this end, in addition to testing the traditional peak metrics, novel analytics were applied to quantitatively assess the time series generated by the instrumented mouthguard.

3.4. Materials and Methods

CSx instrumented mouthguards (CSx, CSx Systems Ltd, Auckland, New Zealand) were compared against reference sensors (REF) over a range of impact conditions in a laboratory setting. First, uniaxial (one-degree-of-freedom, 1DOF) drop tests were conducted, evaluating the accuracy of each axis of the mouthguard's linear accelerometer and angular rate sensor. Second, the mouthguards were affixed to an instrumented headform that was freely dropped onto flat and oblique anvils, generating different combinations of linear and angular kinematics. These six-degree-of-freedom (6DOF) tests were aimed at quantifying the combined performance of all six measurements to more realistic impact conditions. For both groups of tests, comparative analyses were performed between the CSx sensor's output and the reference sensor's output for key metrics and time series data.

Four CSx mouthguards were custom fitted by the manufacturer to upper dental impressions from three different males. Each mouthguard housed a triaxial accelerometer (ADXL375, Analog

Devices, Norwood, MA, linear range ±200 g, sampling frequency 3,200 Hz) and a triaxial angular rate sensor (ITG-3701, Invensense, TDK, San Jose, CA; angular range ±4000 °.s^{-1}, sampling frequency 1,000 Hz, cross-axis sensitivity ±5%). Data were continuously recorded into a circular buffer until any one of the three axes of linear acceleration registered a value above the predetermined threshold of 10 g. When the threshold was crossed, 50 ms of data (5 ms pre-trigger and 45 ms post-trigger) were stored. The stored CSx data were accessed utilising CSx's proprietary app (CSx Headguard v0.28) and the raw signals of all trials (3DOF linear acceleration (LA) and 3DOF angular velocity (AV)) were downloaded for processing.

One degree-of-freedom tests

For the linear tests, each axis of the CSx accelerometer was compared to a reference uniaxial accelerometer (Kistler 8702B500, range ±500 g, sampling frequency 10 kHz). Both devices were mounted on a 4.8 kg rail-guided impactor, falling onto a padded surface (Figure 3.1a, Figure B.3 on page 169). For the angular tests, each axis of the CSx angular rate sensor was compared to a reference uniaxial angular rate sensor (IES 3101-9600, range ±9600 °.s^{-1}, sampling frequency 10 kHz). Both devices were affixed to a rotating platform onto which the same impactor fell (Figure 3.1b). A dental impression assembly was used for both configurations (Figure 3.2a, Figure 3.2b). Different impactor drop heights (0.1 to 0.9 m) and sheets of expanded polystyrene (EPS) padding of different densities were used to vary the impact magnitudes (40-170g) and impact durations (8 to 12 ms) to achieve a range of test conditions. Six conditions, defined as the combination of targeted impact magnitude and duration, were tested for the linear impacts, and seven for the angular impacts; three drops were repeated successively for each test condition (see test matrices in Appendix B, Table B.1 and Table B.5). A single mouthguard sample was used for each test.

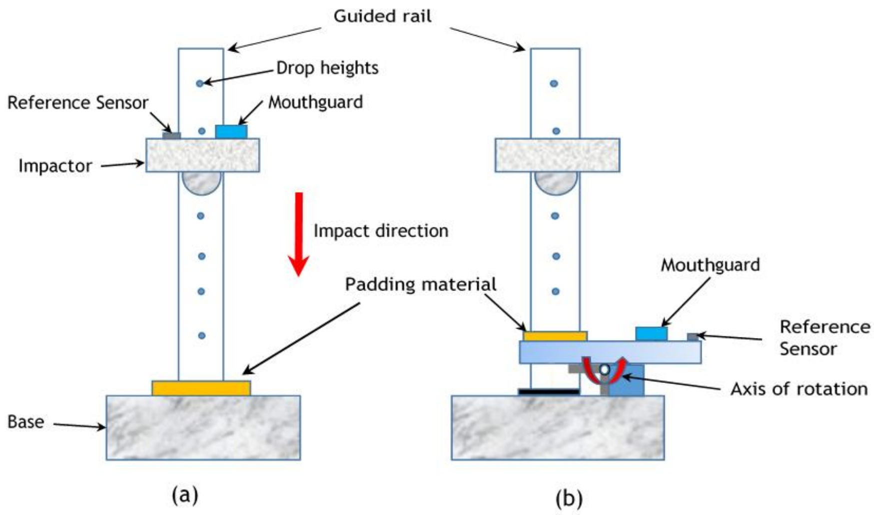


Figure 3.1. Test setup on the guided rail for (a) the linear 1DOF tests and (b) the angular 1DOF tests.

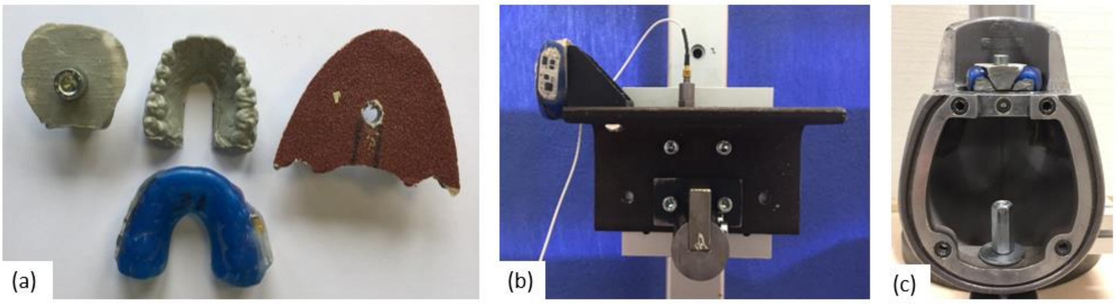


Figure 3.2. (a) Custom-made assembly to attach the mouthguard on the test devices, (b) the mouthguard fixed to the linear impactor for an impact in the +X direction, (c) the mouthguard fixed to the exterior surface of the Hybrid III headform in the vicinity of the palate.

Six degree-of-freedom tests

For the 6DOF tests, a mouthguard was mounted to the inferior surface of a male 50th percentile Hybrid III headform in the vicinity of the palate via the custom-made assembly (Figure 3.2c). The headform was equipped with reference sensors located at its centre of gravity: a triaxial accelerometer (356B21, PCB Piezotronics Inc., range ±500 g) and a triaxial angular rate sensor (AREF-06, ATA sensors, range ±200 rad.s^{-1}). The reference sensors were sampled simultaneously at a frequency of 25,600 Hz. The headform fell freely from various heights (0.4 to 3.5 m) onto flat or 45° anvils covered with EPS sheets of different densities to generate a wide range of linear and angular accelerations (the reference sensor's processed LA and AA resultant peaks (PLA and PAA, respectively) ranged from 10 to 209 g and 662 to 8,869 rad.s^{-2}). Five impact directions were tested: two onto the flat anvil and three onto the oblique anvil (Figure 3.3). Multiple conditions were tested for each impact direction. Drops were repeated successively three times for each condition (Table B.9). Each condition was tested with a single CSx mouthguard.

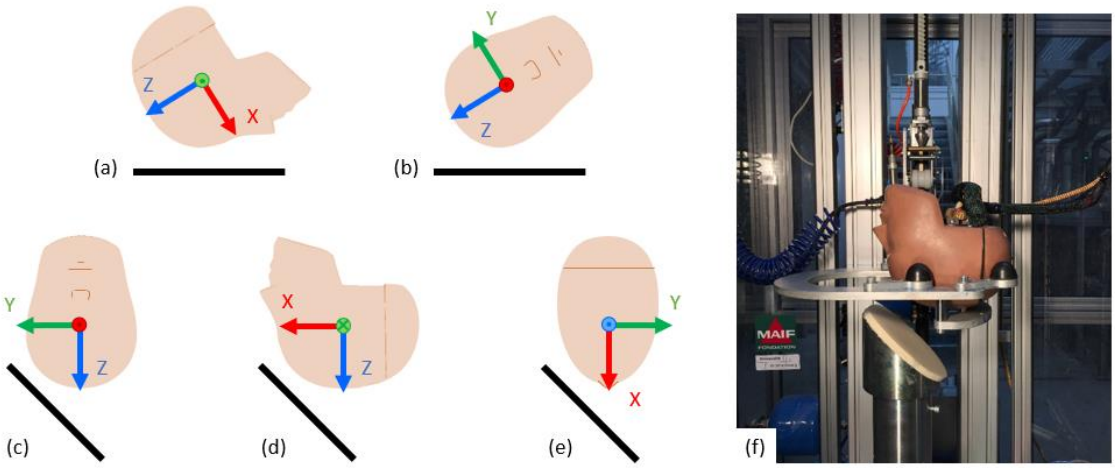


Figure 3.3. Illustration of the drop test setup and headform's orientation for each direction of impact: (a) flat frontal, (b) flat lateral, (c) oblique coronal, (d) oblique sagittal and (e) oblique horizontal. (f) depicts the oblique sagittal test.

Data processing

All raw data from CSx and REF were processed using MatLab (R2016b, MathWorks, Natick, MA). For each REF signal, the mean value of the signal over a 20-ms window before the onset of the pulse was subtracted from each data point to 'zero' the signal. The CSx sensors presented a delay between the onset of the LA and AV signals (Figure B.2), which the manufacturer later confirmed was 12-ms. Therefore, we deleted the first 12 ms of angular data for all CSx recordings to temporally align the LA and AV signals. The time series data from the CSx and REF were filtered using a fourth-order low-pass Butterworth filter with cut-off frequencies of 300 Hz for LA and 150 Hz for AV. The CSx AV signals were then interpolated, while the REF signals were down-sampled, to match the CSx LA sampling frequency (3200 Hz). The orientation of the CSx axes with respect to the reference sensors' axes was the result of a 5° rotation about the headform's Y-axis, followed by a 5° rotation about the new Z-axis. A rotation matrix was used to align CSx's LA and AV with the reference frame for all tests. For the 6DOF tests only, the CSx's AA was calculated using the 5-point stencil differentiation method[57] and its LA was transformed to the headform's centre of gravity (CG) using rigid body kinematics. The vector from the mouthguard's sensor location to the headform's CG was measured with a calliper directly from the assembly and resulted in [-0.0275, 0.026 and 0.0611] meters in the headform's coordinate system. Finally, the CSx and REF signals were aligned in time when the unfiltered dominant (1DOF) or resultant (6DOF) LA signal crossed a 5 g threshold.

For the remainder of the analysis, the dominant signal was utilised for the 1DOF comparisons, and the resultant signals for the 6DOF comparisons. Two discrete metrics were computed from both the REF and CSx sensor data for each test: the peak (maximum absolute value of the signal) and the area under the curve (AUC). The AUC was calculated over a fixed window of 15 ms (LA and AA) or 20 ms (AV) from the crossing of the 5 g threshold.

Statistical analyses

Statistical analyses of the discrete metrics were performed in Matlab and R Studio (R version 3.4.4, R Foundation for Statistical Computing, Vienna, Austria).

A follow-up analysis was conducted on the delay between the linear and angular data measured by the CSx to assess whether the manufacturer's claim of 12 ms was accurate. After aligning the linear data of the CSx and reference sensor, three points were calculated on the rising portion of the CSx AV resultant curve and the corresponding points (same magnitude) on the REF AV resultant curve were identified. The delay between these points was averaged across all 6DOF trials.

Linear models were fit to establish the correlation between CSx and REF for the peak and AUC. Preliminary analyses showed that: (1) the variance of the errors was not constant over the range of values; (2) the direction of impact significantly affected the slopes of the linear fits in many cases; and (3) the number of degrees of freedom was limited for the 1DOF tests. Therefore, robust regression models were applied, using the Tukey bisquare (or byweight) estimator, and setting the threshold for significance at 0.05. Simple models were used for all 1DOF tests, with direction-specific intercepts and a common slope (reported in Appendix B on pages 173, 174, 181, and 182); full models were used for the 6DOF tests, with direction-specific intercepts and slopes (reported in the Results). The spread of the data for the 1DOF area under the LA curve was insufficient to allow for the analysis. The results of the models are reported, with p-values indicating if the intercepts and slopes are significantly different from 0 and 1, respectively. The 95th percentile prediction intervals (95th PI) were calculated and the half-width of the corridor at the overall mean is reported and was used to assess the reliability of the measurements.

Two measures of difference between REF and CSx across multiple impacts were also calculated. The mean absolute error (MAE%, Equation 3.1) and the root mean square error (RMSE%, Equation 3.2) are reported for each direction of impact.

$$MAE\% = \frac{1}{n}\sum_{i=1}^{n}\left(\frac{|REF_i - CSx_i|}{REF_i}\right) \quad \text{Equation 3.1}$$

$$RMSE\% = \sqrt{\frac{1}{n}\sum_{i=1}^{n}\left(\frac{REF_i - CSx_i}{REF_i}\right)^2} \quad \text{Equation 3.2}$$

Validity was achieved if the slope and intercept were not significantly different from 1 and 0,[369] respectively, and the MAE% was below 10%.

Time series analysis

A visual, qualitative analysis was first conducted on the time series for the dominant signal for the 1DOF tests, and resultant for the 6DOF tests. The REF and CSx time series were then quantitatively compared using CORrelation and Analysis (CORA)[141] to further evaluate the CSx mouthguards. The CORA method provides an objective evaluation of the discrepancy between two time series via the combination of several independent and complementary methods. The corridor method quantifies how well an experimental curve fits into corridors around a reference curve. The cross-correlation method is comprised of three sub-methods: a cross-correlation value (i.e., similarity of shape), a time shift between the signals, and a comparison of the areas under the two curves. Our analysis focused on the first 15 ms (LA, AA) and 20 ms (AV) of the signal from the 5 g threshold crossing. Constant corridor widths were automatically calculated by CORA, using 5% and 50% of the REF signal's peak amplitude for the inner and outer corridors, respectively. Each (sub)method provides a rating ranging from 0 (no correlation) to 1 (perfect match) and the final CORA score is the weighted average of the four ratings. For each (sub)method, a quadratic transition was chosen to describe the decline of the rating between 1 and 0. The weights were defined as 0.50 for the corridor method, and 0.167 for each of the cross-correlation, area under the curve and time shift sub-methods.

Of the 234 impact tests conducted, a total of 6 (3%) were excluded from all analyses because the CSx accelerometer saturated. Another 18 (8%) were excluded from the AUC and CORA analyses because the CSx LA signal started over 5 g and could not be aligned in time with the REF signal (n = 14), or there were motion artefacts (n = 3) or electrical artefacts (n = 1) (see Appendix B for more information, on pages 169, 176, and 184 for the 1DOF linear, 1DOF angular and 6DOF tests, respectively).

3.5. Results

Across all impacts, the visual analysis revealed a good level of agreement between the shapes of the REF and CSx time series (Figure 3.4, Figure B.4, Figure B.7, Figure B.10-14). While the shape appeared similar, the magnitude of the signals differed inconsistently across the conditions, for both the individual components and the resultant signals. The CSx angular signals also showed an inconsistently delayed onset when compared to the REF data, despite having corrected for a 12-ms delay that was confirmed by the CSx manufacturer (Figure 3.4). Additional analysis calculated from the raw data revealed an average total delay of 13.93 ms, resulting in a final misalignment of 1.93 ± 0.67 ms (range 0.17-3.68 ms).

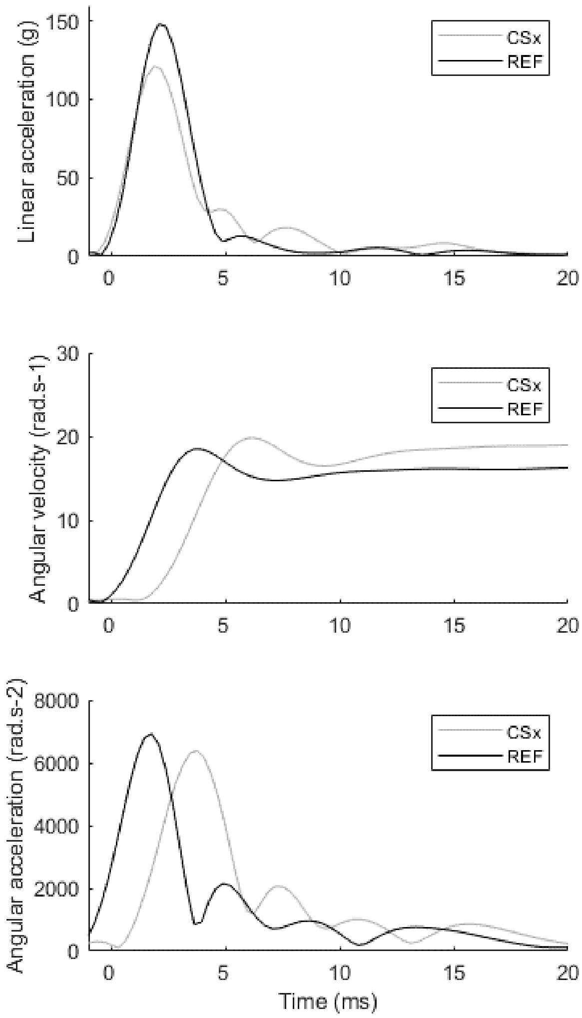


Figure 3.4. Processed resultant linear acceleration, angular velocity and angular acceleration from the CSx mouthguard (gray line) and the reference sensor (black line) from an example 6DOF flat lateral impact, after having corrected for a 12-ms delay.

Only two of the peak kinematic measurements met all three criteria defined to satisfy the assumption of validity (AV in flat frontal and oblique sagittal directions; Figure 3.5, Figure B.15, Table 3.1). The linear regression models all resulted in significant correlations (p-values <0.001) and the variability between the REF and CSx values was well explained by the regressions ($R^2 > 0.88$ for the 1DOF tests, > 0.98 for the 6DOF peak values, and > 0.93 for the 6DOF AUC values). For the peak AV measurements, two directions met our three validity criteria, one other direction met two of the criteria, and the remaining two were close to meeting two of the criteria. Specifically, all intercepts were not significantly different from 0 (-0.832 to 0.748 $rad.s^{-1}$), most slopes were not significantly different from 1 (0.848 to 0.947), and the MAE% was below 10% for two directions of impact (7-20%), with the overall MAE being of 14 ± 8% (range 1-57%). However, no direction of impact met more than one criterion of validity for the peak LA, or more than two criteria for the peak AA. The peak LA and AA intercepts were systematically different from 0 (range for PLA: -11.477 to 3.272 g, PAA: -444.350 to 265.982 $rad.s^{-2}$) and in half the cases the slopes were different from 1 (PLA: 0.812 to 1.221, PAA: 0.787 to 1.252) and/or the MAE for the direction was over 10% (PLA: 8-21%, PAA: 5-13%). The MAEs across all trials averaged 13 ± 8% (range 0-38%) for LA and 8 ± 6% (0-27%) for AA.

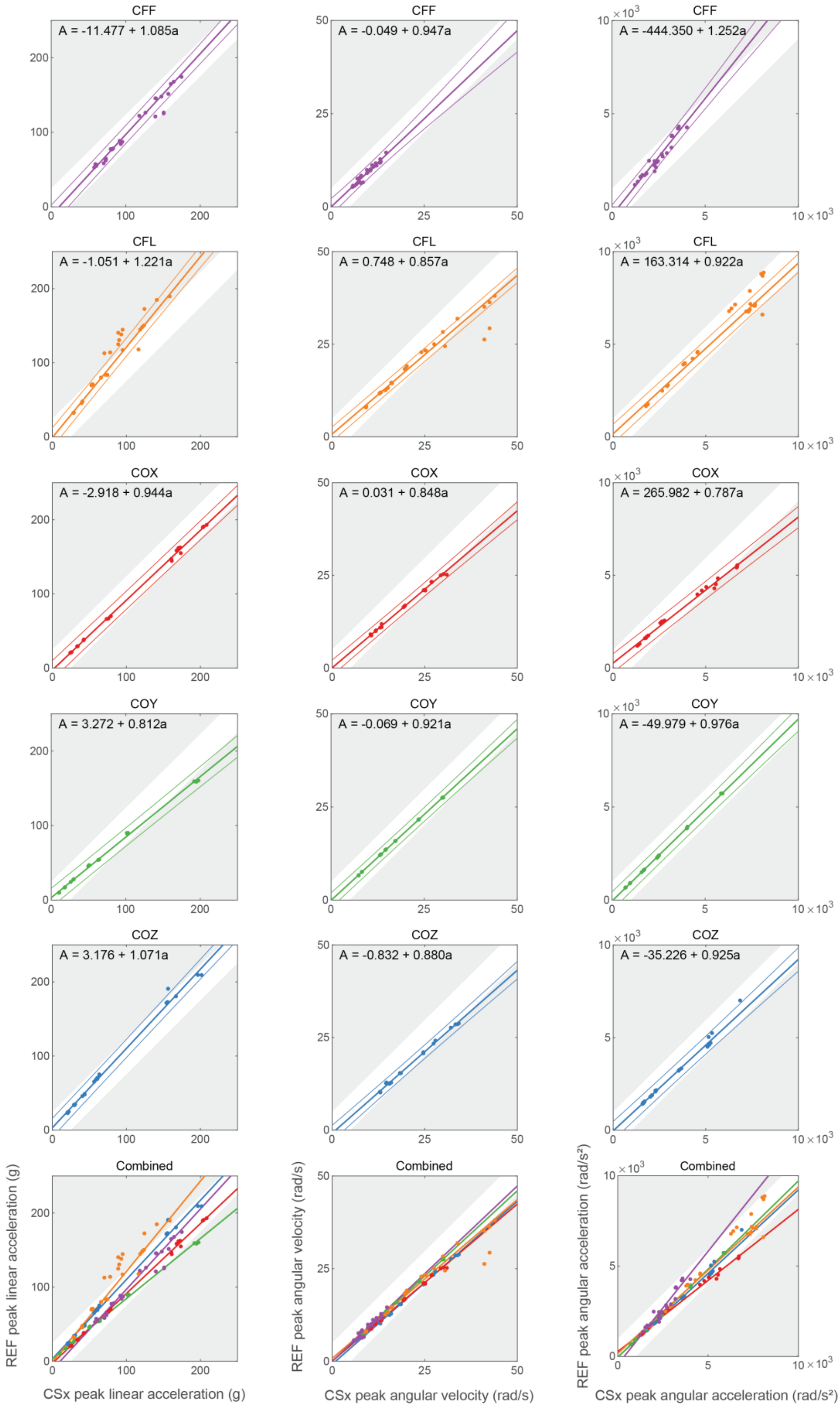


Figure 3.5. Linear regression plots of the peak metric for the linear acceleration, angular velocity and angular acceleration from the 6DOF impacts. CFF: flat frontal; CFL: flat lateral; COX: oblique coronal; COY: oblique sagittal; COZ: oblique horizontal. The bold line represents the slope; the fine lines represent the 95th percentile prediction interval; the shaded area represents ±10% of the plot's full scale about the diagonal. For the equations, "A" represents the reference sensor value and "a" represents the CSx sensor output.

Table 3.1. Intercepts and slopes of the linear regression models and mean errors for the peak linear acceleration, angular velocity and angular acceleration from the 6DOF impacts. Bold values indicate conditions that meet our criteria for validity, i.e., a slope or intercept not significantly different from 1 or 0, respectively, or a MAE% less than 10%.

		Intercepts				Slopes				Errors	
	N	Estimate	SE	tStat	*p**	Estimate	SE	tStat	*p**	MAE%	RMSE%
Peak linear acceleration											
Flat Frontal	26	-11.477	3.475	-3.303	0.001	1.085	0.030	-2.808	0.006	**8%**	10%
Flat lateral	27	-1.051	4.515	2.309	0.023	1.221	0.044	-4.972	<0.001	21%	23%
Oblique coronal	21	-2.918	4.190	2.043	0.044	0.944	0.036	1.585	**0.116**	12%	13%
Oblique sagittal	21	3.272	3.999	3.688	<0.001	0.812	0.037	5.017	<0.001	12%	13%
Oblique horizontal	21	3.176	4.100	3.574	<0.001	1.071	0.037	-1.898	**0.060**	12%	12%
Peak angular velocity											
Flat Frontal	26	-0.049	0.693	-0.070	**0.944**	0.947	0.068	0.775	**0.440**	**7%**	11%
Flat lateral	27	0.748	0.799	0.997	**0.321**	0.857	0.070	2.040	0.044	14%	18%
Oblique coronal	21	0.031	0.890	0.089	**0.929**	0.848	0.073	2.070	0.041	18%	18%
Oblique sagittal	21	-0.069	0.836	-0.024	**0.981**	0.921	0.073	1.072	**0.286**	**9%**	9%
Oblique horizontal	21	-0.832	0.914	-0.857	**0.393**	0.880	0.073	1.639	**0.104**	20%	20%
Peak angular acceleration											
Flat Frontal	26	-444.350	154.064	-2.884	0.005	1.252	0.059	-4.259	<0.001	**6%**	10%
Flat lateral	27	163.314	195.563	3.107	0.002	0.922	0.063	1.254	**0.212**	**6%**	8%
Oblique coronal	21	265.982	185.574	3.828	<0.001	0.787	0.065	3.296	0.001	13%	15%
Oblique sagittal	21	-49.979	179.858	2.193	0.031	0.976	0.066	0.357	**0.722**	**5%**	6%
Oblique horizontal	21	-35.226	190.000	2.153	0.034	0.925	0.067	1.117	**0.267**	**9%**	10%

SE = standard error; MAE% = Mean absolute error; RMSE% = Root mean square error.
*the p-value for the intercept terms indicates whether it is significantly different from 0, while the p-value for the slope terms indicate whether it is significantly different from 1.

The AUC data behaved better than the peak's, with all intercepts not significantly different from 0 and most slopes not significantly different from 1 (Figure B.15, Table B.10). The MAE% averaged 16 ± 10%, 7 ± 7% and 8 ± 6% for the LA, AV and AA AUCs, respectively. As a result, for the AUC metric, all directions of impacts met all three criteria for validity for AA, as well as four directions for AV.

The reliability was assessed with the overall half-width 95th percentile intervals, which resulted in 12.25 g, 1.83 rad.s^{-1}, 470.54 rad.s^{-2} for the peak LA, AV and AA, respectively. These are equivalent to 14, 10 and 13% of the CSx mean values and reflect, for example, that we can be 95% confident that the actual PLA value is within ± 12.25 g of the value measured by the CSx sensor. Additionally, the small differences between the MAE% and the RMSE% within a test series indicated that the dispersion of the errors between CSx and REF was consistent and with few outliers.

The average CORA scores were higher for the LA signals (range of the means by impact direction: 0.76-0.93) than for the AV (0.56-0.68) and AA signals (0.48-0.64)(Table 3.2). The cross-correlation ratings indicated that the shape of the REF and CSx signals was similar (≥ 0.91) while their size differed, as shown by the lower and more variable area under the curve ratings (0.41-0.82). Low time shift (0.30-0.57) and corridor ratings (0.33-0.61) were observed for the angular signals, in comparison to the linear signals (time shift: 0.70-0.97, corridor: 0.79-0.94). The scores for the angular signals were affected by the poor alignment of some CSx AV and AA curves with the reference curves.

Interestingly, the results indicated that the different samples of the CSx sensors exhibited different behaviours. Specifically, for the 1DOF angular tests about the Z-axis, simple linear regressions conducted separately on the data from the two CSx samples used revealed a highly fit model for one mouthguard (n = 12, slope of 0.98 and R^2 of 1.00) and a moderately fit model on the other (n = 8, slope of 0.48 and R^2 of 0.78)(Figure B.8 in page 181).

Table 3.2. Mean (standard deviation) and [range] of the overall CORA scores and sub-methods ratings for the 6DOF tests by direction of impact.

	N	Overall CORA score	Corridor	Cross-correlation	Area under the curve	Time shift
		Weights	*0.50*	*0.167*	*0.167*	*0.167*
Linear acceleration						
Flat Frontal	26	0.93 (0.05) [0.80 - 1.00]	0.94 (0.05) [0.77 - 1.00]	1.00 (0.00) [0.99 - 1.00]	0.79 (0.12) [0.47 - 0.98]	0.97 (0.05) [0.89 - 1.00]
Flat lateral	27	0.78 (0.07) [0.64 - 0.87]	0.79 (0.09) [0.60 - 0.88]	0.99 (0.01) [0.97 - 1.00]	0.41 (0.17) [0.13 - 0.65]	0.94 (0.08) [0.69 - 1.00]
Oblique coronal	19	0.84 (0.05) [0.70 - 0.89]	0.83 (0.05) [0.68 - 0.90]	0.99 (0.01) [0.97 - 1.00]	0.59 (0.05) [0.46 - 0.65]	0.93 (0.10) [0.69 - 1.00]
Oblique sagittal	18	0.84 (0.04) [0.76 - 0.89]	0.86 (0.04) [0.79 - 0.92]	1.00 (0.00) [0.99 - 1.00]	0.57 (0.08) [0.41 - 0.70]	0.92 (0.16) [0.51 - 1.00]
Oblique horizontal	21	0.76 (0.08) [0.66 - 0.92]	0.79 (0.07) [0.73 - 0.93]	0.99 (0.00) [0.98 - 1.00]	0.51 (0.11) [0.38 - 0.82]	0.70 (0.23) [0.37 - 1.00]
Angular velocity						
Flat Frontal	26	0.67 (0.10) [0.47 - 0.86]	0.59 (0.14) [0.34 - 0.85]	0.99 (0.01) [0.97 - 1.00]	0.70 (0.16) [0.35 - 0.93]	0.53 (0.14) [0.4 - 0.94]
Flat lateral	27	0.62 (0.09) [0.36 - 0.78]	0.53 (0.11) [0.18 - 0.73]	0.99 (0.01) [0.97 - 1.00]	0.67 (0.16) [0.21 - 0.98]	0.46 (0.06) [0.4 - 0.52]
Oblique coronal	19	0.56 (0.03) [0.50 - 0.64]	0.46 (0.05) [0.37 - 0.55]	0.99 (0.01) [0.97 - 1.00]	0.57 (0.06) [0.49 - 0.70]	0.45 (0.06) [0.4 - 0.52]
Oblique sagittal	18	0.68 (0.03) [0.63 - 0.72]	0.61 (0.07) [0.48 - 0.72]	0.99 (0.01) [0.98 - 1.00]	0.78 (0.08) [0.68 - 0.94]	0.47 (0.06) [0.4 - 0.52]
Oblique horizontal	21	0.63 (0.05) [0.56 - 0.72]	0.58 (0.10) [0.41 - 0.74]	1.00 (0.00) [0.99 - 1.00]	0.50 (0.09) [0.34 - 0.63]	0.57 (0.14) [0.4 - 0.94]
Angular acceleration						
Flat Frontal	26	0.64 (0.10) [0.50 - 0.94]	0.56 (0.16) [0.29 - 0.95]	0.97 (0.03) [0.87 - 1.00]	0.82 (0.15) [0.35 - 1.00]	0.40 (0.18) [0.24 - 0.89]
Flat lateral	27	0.57 (0.06) [0.49 - 0.67]	0.47 (0.09) [0.31 - 0.62]	0.91 (0.07) [0.73 - 0.98]	0.80 (0.15) [0.43 - 0.99]	0.30 (0.06) [0.24 - 0.38]
Oblique coronal	19	0.48 (0.03) [0.41 - 0.54]	0.33 (0.05) [0.24 - 0.43]	0.93 (0.06) [0.77 - 0.98]	0.65 (0.07) [0.55 - 0.75]	0.30 (0.06) [0.24 - 0.37]
Oblique sagittal	18	0.52 (0.04) [0.44 - 0.58]	0.36 (0.08) [0.22 - 0.50]	0.96 (0.04) [0.90 - 1.00]	0.77 (0.04) [0.72 - 0.84]	0.31 (0.06) [0.24 - 0.38]
Oblique horizontal	21	0.61 (0.09) [0.47 - 0.80]	0.53 (0.15) [0.33 - 0.82]	0.98 (0.03) [0.91 - 1.00]	0.66 (0.05) [0.61 - 0.80]	0.41 (0.12) [0.24 - 0.69]

3.6. Discussion

This study aimed to assess an instrumented mouthguard under various impact conditions. The CSx mouthguards seldom met our criteria for validity for the measurement of peak kinematics. Indeed, despite robust correlations between CSx and REF (R^2 > 0.98), few 6DOF peaks were measured with enough accuracy that the slope and intercept of the linear model were not different from one and zero, respectively, and that the mean absolute errors were below 10%. The CORA analysis of the time series data revealed excellent agreement in terms of signal shape between CSx and REF and identified differences between CSx and REF in terms of magnitude and timing, especially in the angular data.

Overall, the linear acceleration peaks and AUCs were not estimated with enough accuracy to satisfy our definition of validity. The linear regression and CORA analyses showed that the signals measured by the CSx sensors either under- or over-estimated the reference signals. The direction of the errors varied across the impact conditions and the mouthguard samples, suggesting that the measurements may be affected by the geometrical and physical properties of the mouthguards. The mouthguards' custom moulding process leads to variability in the position and orientation of the sensors between the layers of material and in the general shape of the mouthguard. Testing a larger number of mouthguard samples, and investigating the accuracy of the signals along the three axes of measurement rather than the resultant signal may help understand the source of these errors.

The results also showed that the absolute errors between CSx and REF measurements increased with increasing impact magnitude, for both the peak (MAE%: 8-21%) and the AUC (8-26%). Nonetheless, the errors produced by the CSx mouthguards were comparable to other head impact systems, such as the X2 mouthguard (pooled PLA MAE%: 13% for CSx, 12% for X2)[369] or the G-Force Tracker after adjustment using regressions (PLA: 3.5-23.0%,).[3] The CSx mouthguard also performed better than other systems used in previous studies, such as the HIT System (PLA: 32%, including facemask impacts)[369] and the xPatch (PLA: 31.1%).[363] Despite being comparable or better than other devices, our findings imply that caution is needed when reporting and interpreting peak linear accelerations and presumably also for severity metrics calculated using the integral of the linear acceleration curve, such as the Head Injury Criterion.[402] While the analysis of individual impacts is discouraged, the reporting of population-wide summaries (e.g., the number of impacts by bins of varying magnitude) would be acceptable.

Angular kinematics were measured more accurately than the linear kinematics. Specifically, the angular velocity and acceleration AUCs generally met our validity criteria, and therefore, these metrics, or other metrics calculated using these measurements can be reported and interpreted with confidence. However, the peak angular acceleration results were comparable to the linear acceleration findings, where variations occurred across directions of impact, resulting in general heteroskedasticity. Despite showing only limited accuracy, the CSx performed again in a better or comparable way to the X2 mouthguard (pooled PAA MAE%: 8% for CSx, 41% for X2),[369] the HIT System (PAA: 35% including facemask impacts),[369] or the xPatch (PAA: 18.9%).[363] The peak angular velocity was measured by CSx with good level of validity, as indicated by the intercepts and slopes, however the large MAE% (7-20%, pooled MAE%: 13%) showed limited accuracy. Similar accuracy has been reported for the G-Force Tracker after adjustment using regressions (PAV: 2.1-18.8% for G-Force Tracker),[3] and better accuracy has been measured for the xPatch (PAV: 1.8%).[363]

The overall CORA scores ranged from poor to excellent (range 0.36-1.00), with the LA scores being generally higher than the AV and AA scores. Corroborating the visual and regression analyses, sub-method ratings confirmed that the shape of the signals was well matched for all kinematic measurements (cross-correlation ratings from 0.73 to 1.00), whereas the other ratings were more variable and reflected issues with the CSx sensors' measurement (corridor ratings: 0.18-1.00, area under the curve: 0.13-1.00, time-shift: 0.15-1.00). Because of the apparent delayed onset for the angular data, the results from any analysis that combines angular and linear data on a similar timeline without further processing (e.g., the transformation of linear acceleration to the head's CG, or numerical modelling of brain strains), needs to be carefully interpreted. Furthermore, time-related information from the angular time series, such as the time of onset, or time-to-peak, should be used cautiously.

Although the peak value is the most reported metric in head impact literature, a visual comparison of the REF and CSx curves shows that the peak value alone cannot adequately characterize the differences between the sensors. Because some severity metrics utilise the integrals of the kinematic signals, we assessed the AUC as an additional indicator of the validity of a device's measurement. Ultimately, if sensor data are to be used as inputs to computational brain models, then the shape of the head kinematic signals may be as important as the peak magnitude when calculating brain strains and inferring injury potential.[121, 182, 212] Additionally, with head impact devices, such as the CSx mouthguard, now allowing the user to obtain time-series data, a shift towards more comprehensive curve analyses can be expected.[226] Complementing the conventional approach of assessing peak values alone, the analysis of the time series may be an important step toward a more meaningful method for evaluating head impact sensors. In the context of sensor use *in-vivo*, this dual approach would allow the user to know which aspects to trust (in the case of the CSx mouthguard, the general shape of the pulse), which to interpret with caution (i.e., the magnitude and LA AUC), and which to question (i.e., angular kinematics time-related information).

Our testing revealed two major shortcomings of the CSx mouthguard. Firstly, despite correcting for a systematic time shift, the signals were still poorly aligned. This may have introduced errors in the calculation of the LA at the headform's CG and would be particularly problematic if the time series were used as input to computational brain models. Secondly, we observed different behaviours between sensor samples during the 1DOF tests. Variations in the way the sensors are moulded into the mouthguard may explain the differences, although other experimental or product-related issues cannot be excluded. In contrast to these shortcomings, the CSx system has the advantage of providing the end user with the raw data. Obtaining the raw data enables a custom analysis, and in this study, allowed us to identify and partly correct for the temporal misalignment. We were also able to choose and report our processing parameters, something which may not be possible for users of "black box" systems. Data accessibility is important to improve transparency and standardisation of methods across studies.

Finally, this study is not without limitations. The impact conditions tested in this study were not based on a specific sport. Because sports present different distributions in impact location, which can influence a sensor's accuracy,[232] the overall errors reported may not be representative of any sport. The results from the 1DOF tests must be interpreted with caution as the alignment between the measuring axes of the mouthguard and the reference sensor could not be verified. Also, the discrepancies between different samples of the CSx mouthguard could not be investigated further because of the low number of samples available and their relative lack of robustness. Nonetheless, such discrepancies raise the issue of sample-to-sample variations, which should be evaluated in future studies.

3.7. Conclusion

This study has found that the CSx mouthguards were valid for some measures of head kinematics in response to impact in a laboratory setting. Caution is recommended when reporting and analysing linear acceleration magnitudes and AUCs, and angular velocity and acceleration magnitudes. Despite performing similarly or better than other head impact systems, only the area under the angular velocity and acceleration curves was determined to be valid. The use of CORA complemented the analysis of the discrete metrics and supported the visual and regression analyses. Despite its limitations, the CSx mouthguard may contribute to increasing our understanding of head impacts in sports by allowing the reporting of population-wide summary metrics where the magnitude of individual impacts is not of primary concern.

Chapter 4 - Performance of head impact sensors to record video-verified head impacts in boxing sparring

This manuscript is currently in preparation for journal submission.

Elements of protocol and supplementary materials can be found in Appendix D and Appendix E, respectively.

4.1. Prelude

Instrumented mouthguards have generally demonstrated moderate to good validity under controlled conditions in the laboratory, and the CSx mouthguards were no exception. However, laboratory tests are known to have limited biofidelity at the interface between the headform and the sensor, and studies have highlighted skull/sensor decoupling issues when the sensors were attached to human heads. Moreover, many studies reviewed in Chapter 2 have reported large proportions of false positive events, i.e., acceleration events recorded by a sensor that could not be matched with a head impact observable on video. Therefore, the performance of a given sensor in the laboratory may not be representative of its performance on the field, and Chapter 4 was designed to assess the capacity of the CSx mouthguard to record head impacts *in-vivo*.

To fully apprehend the performance of the mouthguard, it appeared necessary to find a sporting situation that would allow the verification of most, if not all, acceleration events. One such situation occurs in combat sports, which typically involve only two participants in a well-defined, small area. Unfortunately, CSx Ltd. did not support the use of their mouthguard in combat sports, as they had not conducted enough safety tests on the battery and the mouthguard's coating to ensure their integrity, would a punch land directly on it. CSx Ltd. used to also manufacture skin patches, that embarked the same electronic components and used the same algorithms as the mouthguards, and we had the opportunity to obtain some patches. However, because particularly large measurement errors have been reported for skin patches when adhered to human skin, we chose to also obtain instrumented mouthguards from Prevent Biometrics and to simultaneously assess the performance of both types of sensors.

4.2. Abstract

Background: Various types of inertial sensors have been used to quantify the exposure to head impacts, but recent work has suggested moderate accuracy in the numbers measured.

Objectives: (1) Compare the number of acceleration events recorded by various sensors; (2) assess whether the type or location of impact affects the capacity of each sensor to be triggered; (3) evaluate the accuracy of the sensors to record events of particular interest.

Methods: This observational cohort study was designed to record head acceleration events during boxing sparring. Seven amateur boxers were equipped with three sensors simultaneously: an instrumented mouthguard, a skin patch, and a headgear patch, all set with a 10-g trigger. Every contact to the participants' head was identified and characterised from video and compared to the sensors' data. Additionally, a certified boxing judge identified scoring punches from a sample of the video data.

Results: All three sensors showed positive predictive values for head impacts over 96%. Between 49 and 78% of video events did not generate a sensor recording. All three sensors were more likely to be triggered by impacts landing close to the sensor. Skin and headgear patches showed higher sensitivity than the mouthguard in general, and specifically for scoring punches (86 and 78% vs 35%), but lower specificity (76 and 75% vs 90%).

Discussion: Many missed events may have been below the trigger threshold and are likely of little interest relative to injury risks. Although the mouthguard was the least affected, all sensors were more sensitive to impacts occurring in their proximity, suggesting that all sensors may present varying degrees of decoupling from the skull.

Conclusions: There were differences between the sensors, and investigation of the kinematics is needed to advise on the sensors' potential for measuring exposure and informing injury risks.

4.3. Introduction

Numerous devices are being used to record and quantify impacts to athletes' heads during sports participation. The information gained from these devices has been used to evolve rules,[264, 304] adapt practices,[10, 49] or influence the use of protective equipment,[384] all aimed at increasing athlete safety. Injury risk curves have been proposed to describe the relationship between concussion and head impact dynamics.[347] Several studies have also evaluated the association between head impacts and physiological, neurological or cognitive detriments.[280, 361] Many of these studies have connected clinical outcomes to the number of head impacts recorded by head impact devices, but recent work suggests the numbers are less accurate than previously thought.[67, 86, 220, 224, 258, 339, 403]

Among the first reports of spurious recordings - or false positive sensor events, i.e., when a sensor records an acceleration event, but the video review does not show any impact occurring - was the American football study by Hernandez et al. in 2015.[164] The authors reported that 99% of all mouthguard measurements collected were spurious recordings. Cortes et al. later found that most spurious events collected in lacrosse were happening outside of playing periods (e.g., time-outs or player on the sideline),[86] although 25 to 64% of events that occurred during play were not associated with an impact being observed on video. Across several studies, the proportion of false positives and events that could not be video-verified (e.g., player not in view) ranged from 31% to 98%.[5, 67, 86, 220, 224, 322, 403] There also exist several studies reporting false negative events, where an impact to the head is visible on video but the sensor did not record anything. Identifying all true contact periods and/or individual impacts from video is time-consuming and has only been completed in a few studies.[67, 136, 220, 258, 339] In these studies, the false negative rate varied between 26 and 59%, and events of particular interest such as impacts directly leading to the athlete displaying signs of concussion, were missed by the sensors.[199] Two-way video verification, where sensor recordings are matched to video events and video events are matched to sensor recordings, is an important step in the full assessment of a sensor's performance.[220, 258]

Head impact devices record an event when acceleration reaches a threshold. However, actions such as jumping, running, or sitting down, have been shown to trigger spurious recordings in sensors attached to the skin or to headbands.[224, 322, 339] Such sensors, loosely coupled to the skull via soft tissue (the skin) or an external object (headband, helmet), can move independently from the skull.[180, 418] This relative motion affects the magnitude of the acceleration signal, triggering spurious recordings and resulting in overestimates of exposure.[367, 418] Instrumented mouthguards have been shown to be more tightly coupled with the skull[418] but still record false positive events, with spitting or chewing as possible causes.[124] Studies conducted with several sensors simultaneously have been performed in the laboratory but have not reported the number of events accurately recorded[228, 418] or have not included mouthguards.[94, 227] Therefore, it remains unclear whether there is a type of head impact sensor that is more accurate than others.

Most *in-vivo* studies involving one sensor have been hindered by a high proportion of events that could not be verified because the athlete was out of view of the cameras, the quality of the video was low, or the view of the head was obstructed.[136, 416] An imprecise quantification of exposure to head impacts may hinder the advancement of our understanding of concussive injury.[124, 308, 416] In order to help develop more trustworthy devices, the factors that can result in false positives or false negatives need to be better understood.

Thus, this study was designed to record head acceleration events using three sensors simultaneously, representing three types of coupling to the skull: an instrumented mouthguard, a patch attached to the skin and a patch attached to the headgear. The study took place in an observational *in-vivo* environment: boxing sparring. The objectives were to: (1) compare the three sensors in terms of the number of acceleration events recorded relative to events observed on video; (2) assess whether the type or location of impact, or the participant, affects the capacity of each sensor to be triggered; (3) evaluate the accuracy of the sensors to record events of particular interest, in this case, head impacts that a boxing judge would look for in terms of scoring, or that boxers would report as an impact that shook them.

4.4. Methods

Overview

This study is an observational cohort study. A group of competitive amateur boxers were observed during sparring sessions from September to December 2020 in Auckland, New Zealand. Each participant was equipped with an instrumented mouthguard, a skin patch, and a headgear patch. Sparring was filmed using five cameras surrounding two sparring areas. Videos were reviewed to identify every punch or contact to the participants' head and body. Acceleration events recorded by the sensors were then matched to video events and the number of true and false positives were analysed. Additionally, a certified boxing judge reviewed a sample of the video data, identifying scoring punches, i.e., punches landing with good technique and a certain amount of force.

This study was approved by the Auckland University of Technology Ethical Committee (AUTEC 20/153). Data collection and reporting followed the recommendations published by the National Institute of Neurological Disorders and Stroke in the Common Data Elements on video device confirmation[288] and head kinematics estimates.[287]

Definitions

Sparring is defined as "*a form of training involving two boxers in the ring, usually under the supervision of their trainer. Competitive boxers regularly participate in sparring. The aim of sparring is not to win, but to learn and practice, therefore the intensity of a sparring session is typically less than a fight, and the risks are minimised.*"[377] In this study, we primarily observed competitive sparring, except for one session of technical sparring where the boxers focused on specific techniques, e.g., defined punches combinations.

Sparring, like boxing bouts, is organised in <u>rounds</u> interspersed with breaks (Figure 4.1). In the present study, the duration of the rounds was 3 minutes for competitive sparring and 3.5 minutes for technical sparring, with 30-s rest periods. A combination of several rounds, usually three, against the same sparring partner is called a sparring <u>bout</u> in this study. Three minutes of rest (one round) were typically observed between sparring bouts. Finally, a sparring <u>session</u> is defined as the sum of all the rounds performed on one day, i.e., the full duration of a day's sparring action.

The sparring sessions typically lasted 40 to 60 minutes, depending on how many rounds the boxers partook in.

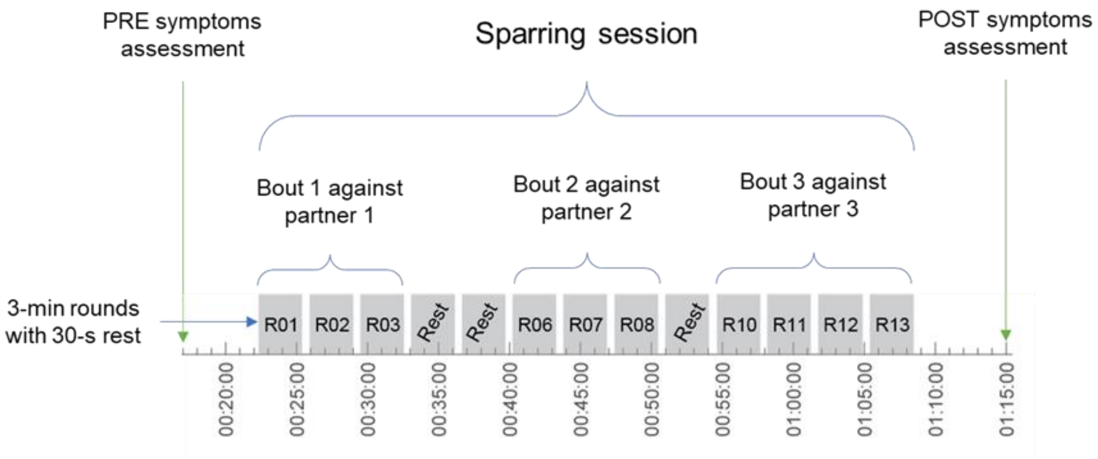


Figure 4.1. Example schedule of a sparring session. The pre- and post-session symptoms assessments times (PRE and POST, respectively) are shown illustratively; refer to Chapter 6, Self-reported concussion history and symptoms, for further detail.

Participants

Seven boxers participated in the study: four females (mean age: 24.0 ± 3.4 years, mean weight: 58.7 ± 6.0 kg) and three males (22.3 ± 2.1 years, 92.3 ± 21.8 kg). Participants were identified by the number of their mouthguard (MG03 to MG10, with MG01 and MG02 kept as back-up). An 8th boxer (MG06) was recruited but withdrew prior to participation due to injury. All participants were amateur boxers (in opposition to professional) and trained under the International Boxing Association (AIBA) rules. All but one were competitive boxers with fighting experience; all were regularly taking part in 1 to 3 sparring sessions weekly. Experience in boxing ranged from 1 to 8 years, and three females and two males had competed at the national level or higher. No participant reported a history of diagnosed concussions.

The following information was collected for each round by the primary investigator: whether the participants were sparring or resting, the area where the sparring took place, the sparring partners, and patch-related information (identification number and patch-related issues). All participants wore headgear during sparring. No participant sought medical attention for a head injury over the course of the study.

Video setup

The participants sparred in one of two areas: a boxing ring and a defined area on the floor. With five cameras, three angles of video were always available for each area (Figure 4.2). Various models of GoPro action cameras (from Hero 3+ to Hero 7 Black) were set up to record at 60 fps with a shutter speed of 1/120 s to minimize motion blur, and at a resolution of 1080p. While a narrower field of view was preferred to avoid image distortion (fisheye effect), some cameras had to be set up with a super-wide field of view ("SuperView") due to space limitations.

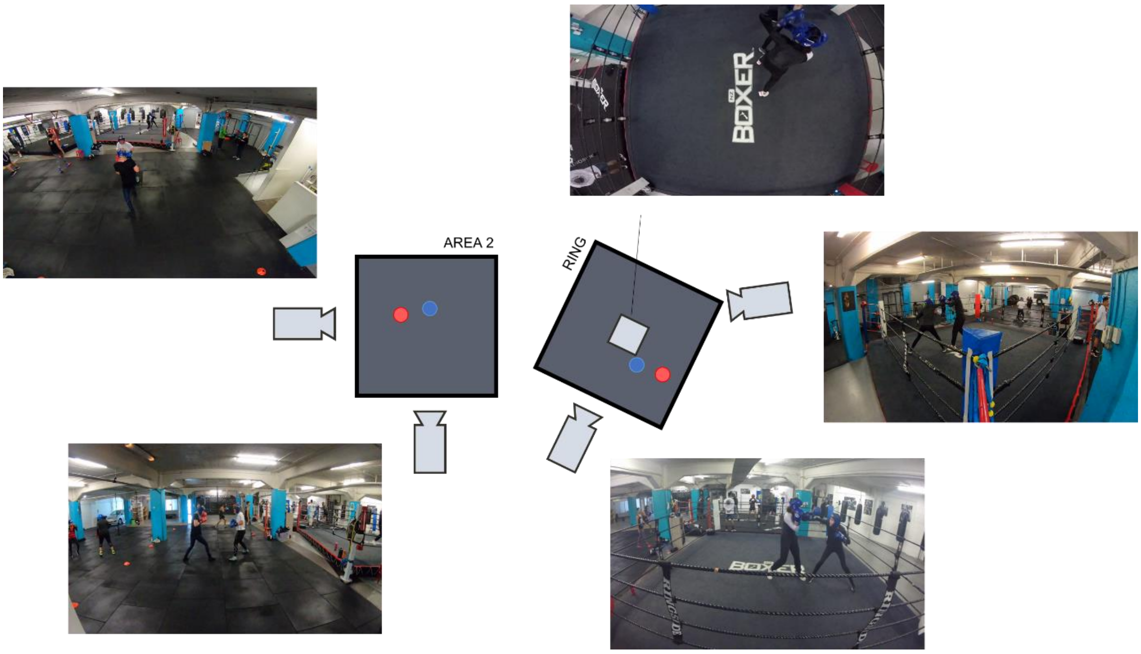


Figure 4.2. Video setup around the two sparring areas. With five cameras viewing two areas, this setup allowed for at least three angles of video for each area.

To ensure that the video and sensors data could be aligned in time, a calibration impact was conducted at the beginning of the session. All sensors were placed on a stiff board that was dropped once in the view of a camera designated as the master camera. This camera was on for the full duration of the session. Later, all videos were synchronised to the master camera and the timestamps of the sensors re-calculated to match the video time.

Instrumentation

The Prevent Biometrics Hybrid mouthguards (Prevent Biometrics Inc., Edina, MN) are boil-and-bite mouthpieces, incorporating a triaxial linear accelerometer (ADXL372, Analog Devices, Boston MA, range ± 200 g) and a triaxial angular rate sensor (BMG250, Bosch, Gerlingen Germany, range ± 35 rad/s), both sampling at 3200 Hz (Figure 4.3A). The sensors are located in the vicinity of the first left lateral incisor.[232] During use, any time the linear acceleration signal reaches 10 g on any axis, the system records a 50 ms acceleration event (10 ms pre- and 40 ms post-trigger) and the recording is transmitted wirelessly to the Prevent Biometrics iOS app. Each recording is timestamped using the iPad clock with a 1-ms resolution. The processed 3-degrees of freedom (3DOF) time series for linear velocity (LV), linear acceleration (LA), angular velocity (AV), and angular acceleration (AA) are then available to download by the end-user. The Hybrid mouthguards used in this study were the latest model from Prevent Biometrics: a boil-and-bite mouthguard that "*replicates the fit and comfort of a custom mouthguard*" (communication with Prevent Biometrics). All Prevent Biometrics mouthguards (the Hybrid, the custom-fit, and the original stiffer and bulkier boil-and-bite) were advertised to incorporate the same hardware and software. The previous models have been shown to identify head impacts with a positive predictive value of 81.6% for the boil-and-bite version in American football (helmeted condition) and 96.4% for the custom mouthguard in rugby (unhelmeted condition).[198]

The CSx patches (CSx Systems Ltd, Auckland, New Zealand) used for this study measure 25 x 21 x 6 mm and weigh 4.5 g (Figure 4.3A). The patches house a triaxial linear accelerometer (ADXL375, Analog Devices, Norwood, MA, linear range ±200 g, sampling frequency 3,200 Hz) and a triaxial angular rate sensor (ITG-3701, Invensense, TDK, San Jose, CA; angular range ±4000°.s^{-1}, sampling frequency 8,000 Hz). A 10-g threshold on any linear acceleration axis was also used. Each recording is 50 ms-long (5 ms pre- and 45 ms post-trigger) and timestamped with a resolution of 1 second using the device's GPS-synched clock. The recordings were stored in the patches' internal memory, which can hold 108 recordings, and the raw 3DOF linear acceleration and angular velocity signals were available via direct download at the end of each session.

Each participant was assigned a mouthguard that was moulded to their upper jaw dentition according to the manufacturer's guidelines (see page 227 in Appendix D). All mouthguards seemingly worked and collected data without interruption or issue. One CSx patch was taped behind the right ear on the flat aspect of the mastoid process (Figure 4.3B). Sports adhesive spray and wig tape (Walker Tape Ultra Hold, West Jordan, UT) were used. We note that the skin patch did in some instances still detach due to intense sweating. A second patch was taped to the back of the headgear using standard double-sided tape (Figure 4.3C). Patches had to be changed between sparring bouts as they would reach their storage capacity. All sensors were thoroughly cleaned, disinfected, and assessed for any sign of damage before and after each session.

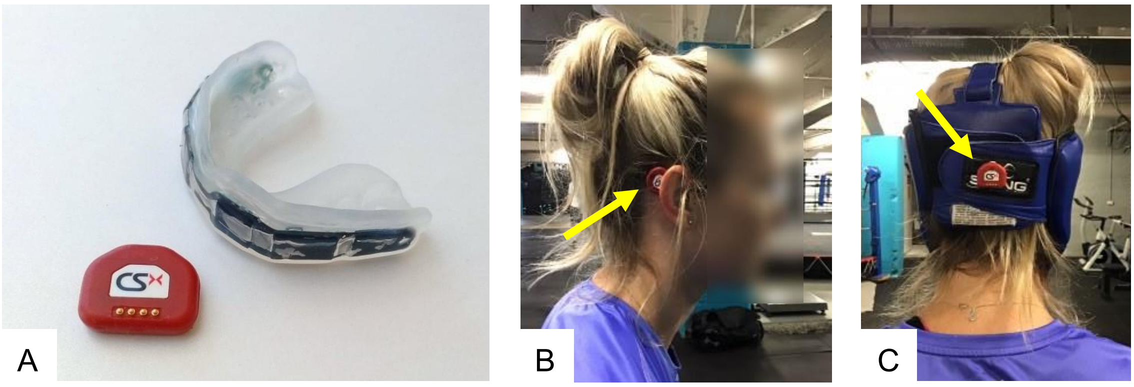


Figure 4.3. A) A CSx patch (bottom-left) and a new, unmoulded Prevent Biometrics Hybrid mouthguard (top-right). B) Placement of the CSx patch on the flat aspect of the right mastoid process (indicated by the yellow arrow). C) Placement of the CSx patch at the back of the participant's headgear (indicated by the yellow arrow).

The Prevent Biometrics mouthguards and the CSx patches both incorporate a proprietary classification algorithm that determines whether the acceleration event is a true head impact or a spurious recording. All recordings were included in this study, independent of the classification. The performance of the algorithms was assessed separately and summarized in Appendix F.

Video processing

Video coding and verification were carried out using Nacsport Elite 6.0.0 software (Nacsport, Canary Islands, Spain). The master video was first used to identify the calibration impact timestamp and extract the start and end times of each round.

The rounds where a participant's sensors collected data from start to end were included for all subsequent analyses. For each of these rounds, all video angles were synchronised in Nacsport and a single coder (ELF) carried out the identification and characterisation of all contact events, watching the videos at about 30% of their normal speed, and viewing each event several times from all angles. The definition of each contact event was discussed and agreed upon beforehand between the primary coder (ELF) and a combat sports expert (SL)(Table 4.1, refer to section Analysis, Intra/inter-rater below). All head-related events were further defined as direct or indirect[387] (Table 4.1, Figure 4.4). A highly inclusive approach was followed, where every contact event was coded, independent of its observed magnitude. The coder used objective criteria, such as the deformation of the glove or a sudden change in the head and/or the punching glove's velocity, to identify contact events. The coder could not be fully blinded to the CSx patches' behaviour as they emit a red light when triggered, but tried to not take it into account when deciding whether there was an event or not.

Table 4.1. Definition of the contact events characterized from the videos.

Type of event	Definition
Definite head impact	Every time the head or headgear is visibly hit or touched.
Head – prolonged contact	When an arm or glove lingers against the head of the opponent, potentially pushing it and rubbing against the headgear; also called "framing" in boxing.
Probable head impact	Every time a head impact is suspected but the video does not allow to be certain (e.g., no good angle, obstruction).
Body impact	Every time the body is visibly hit.
Probable body impact	Every time a body impact is suspected but the video does not allow to be certain (e.g., no good angle, obstruction, loose clothing).
Clinch	Event where a boxer holds the opponent's body and/or arms with one or both of their arms to prevent or hinder the opponent's punches or movements.[387] This may cause the heads to hit or rub against each other.
Artefact	Any event where the sensor may record something not related to a head impact (e.g., the boxer pushes the headgear back in place, or hits the headgear patch at the back of his/her head).
Direct impact	A hit that visibly lands on the head with the knuckle part of the glove without any obstruction and bounces off in the opposite direction.
Indirect impact	A hit that is partially or fully blocked (contact to the head with either boxer's glove or arm), partly deflected or glancing.

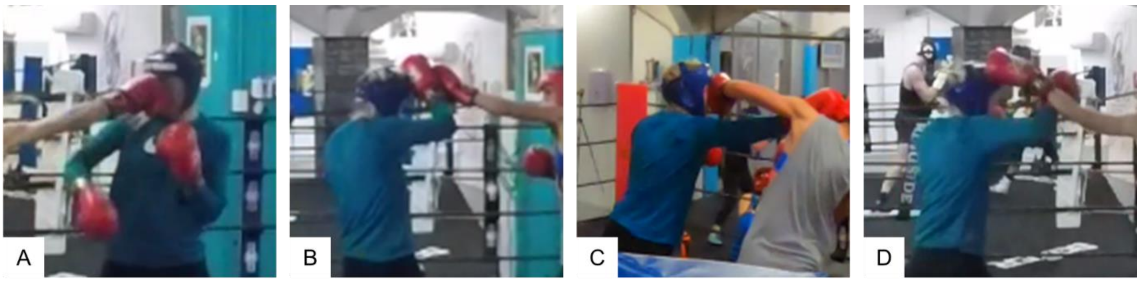


Figure 4.4. Examples of head impacts observed on video. A) Direct head impact, landing cleanly with the knuckle part of the glove, without obstruction. B) Indirect head impact, with contact to the head made by the participant's own glove. C) Indirect head impact from the palm side of the glove, with the punch glancing about the face. D) Indirect head impact, with the punch being deflected by the participant's arm before making contact.

Additionally, each head impact location was manually identified on an image of a helmeted head (Figure 4.5), and this location was later used to calculate the distance (in pixels on the image) between the impact and the estimated location of each sensor. The vicinity of a sensor was defined as an area of 100-pixel radius around this theoretical location, to mimic the size of the knuckle area of a boxing glove (10-15 cm in diameter, depending on the glove).

When a full round was coded, the timeline containing the timestamp (17-ms resolution) and details for each event was exported to a spreadsheet.

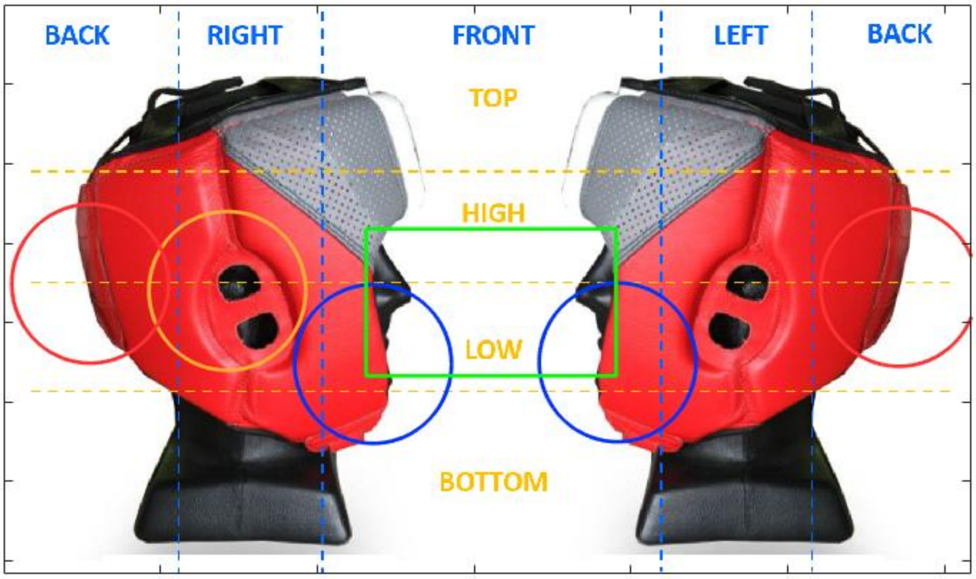


Figure 4.5. Image used on Nacsport to manually point at the location of contact to the participants' head (the same image was used for all participants). Pointing at the image would automatically categorize the impact in a location bin (based on the bins defined by Greenwald et al.[147]) and record the coordinates of the point. The blue, orange and red circles represent the area around the mouthguard, skin patch, and headgear patch, respectively, with an arbitrarily defined radius to mimic the size of the knuckle part of a boxing glove. The green rectangle represents the face area, where for most participants, a punch would not hit the headgear.

Sensor data processing

Sensor data were processed using custom MatLab scripts (R2019a, MathWorks, Natick, MA). The calibration impacts were identified among the sensors' recordings and used to re-calculate the timestamps on the video timeline. The rounds' start and end times were used to automatically classify the sensors recordings based on whether they occurred during sparring action or not. As the kinematics were not used in this chapter, the full methods for the CSx data processing are described in Chapter 5. Of note, 13 skin and 13 headgear patch recordings were excluded because of the incompleteness of the kinematic traces (0.6% of all patch recordings). The remaining recordings had their timestamps exported to a spreadsheet to be matched to the video events.

Video verification

Using Excel (Version 2103, Microsoft, Redmond, WA), the first step of video verification consisted of aligning the sensors' recordings, independent of the video events. This task was performed manually, as despite having re-calculated the timestamps from the calibration impact, the timestamps could differ by up to 4 seconds between sensors (Figure 4.6). It was a known issue, from previous field experience and communication with the manufacturer, that the CSx patches' internal clock could drift after the calibration impact. This drift is in part due to the lack of a GPS signal while inside the gym, so the clock could not re-calibrate itself. The drift was not consistent across sensors. However, once the delay between the mouthguard, the skin patch, and the headgear patch had been estimated using several isolated events, matching every recording was uncomplicated (Figure 4.6).

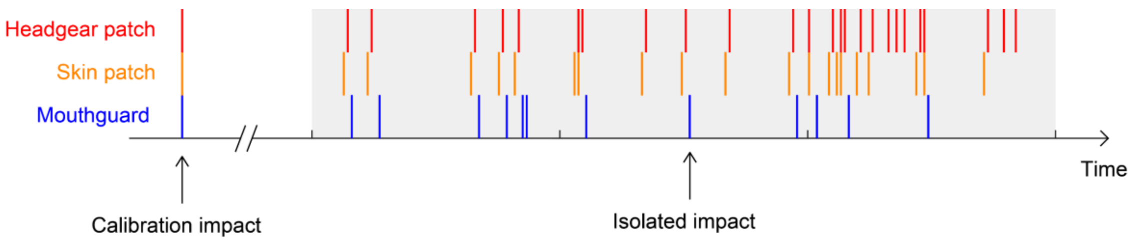


Figure 4.6. Representation of the mouthguard, skin patch, and headgear patch timestamps after alignment on the calibration impact. The timestamps showed the constant delay between the three sensors, which was calculated after having identified several isolated events that could be matched across all three sensors. For example, the highlighted isolated impact had a timestamp of 42:45.856 (min:sec.ms) from the mouthguard, 42:43.000 from the skin patch, and 42:44.000 from the headgear patch. The events on both sides of this isolated impact were approximately 10 seconds away. The grey shaded area represents the 3-minute round, which started approximately 38 minutes after the calibration impact. Note: for ease of understanding, the information relating to the various tools has been colour-coded throughout the figures: mouthguard in blue, skin patch in orange, headgear patch in red, video in green.

The second step of video verification consisted of scanning the video timeline in Nacsport and identifying events occurring within ± 2 s of the sensors' timestamps, accounting for the patches drift and for the fact that the mouthguards' timestamps were the most accurate (± 1.5 s from the video event timestamp). The video events served as reference and if a sensor recording could be matched with a video event, it was marked as a true positive (TP). If a recording existed but could not be matched with a video event, it was marked as a false positive (FP). The video events not associated with an acceleration event were marked as false negative (FN). The criteria for video-verified impacts previously established by Cortes et al.[86] were adapted for this study: (1) the video setup ensured that the participants were visible on video at all times; (2) we differentiated between impacts and probable impacts when we couldn't be certain of the contact to the head; and (3) because boxers can receive several punches to the head in one second (i.e., the patches' timestamp resolution), we marked the sensors recordings where we couldn't identify with certainty which video event they were associated with. The red light emitted by the CSx patches upon trigger was sometimes used as an indicator of which event triggered the sensor when several events happened within 1 s.

Analyses

The details of all video events were loaded into Matlab and combined with the sensor data. Matlab and R Studio (version 1.2.5033, R Foundation for Statistical Computing, Vienna, Austria) were used for analysis.

Intra/inter-rater analyses

Intra- and inter-rater analyses of the video coding step were conducted using two independent coders. The primary coder (ELF) had no previous experience of video analysis and limited experience of boxing but developed the coding template and spent time using the video analysis software. The second coder (SL) is an expert in combat sports, has experience in skill coaching and sport science, but had limited experience with video analysis, and none with the software and template. The goal of this analysis was to determine if a coder without specialist knowledge of the sport could accurately (using a boxing expert as reference) and reliably identify the events of interest using the Nacsport software and custom coding template.

Ten rounds were randomly selected and the first minute of each round was used for the analysis. Each coder reviewed the 1-min segments on two occasions separated by a minimum of three weeks. The results from each assessment were matched and the videos of the missing events were reviewed by the primary coder and marked as either "obvious" when it was a clear miss or "probable" if the contact event could not be strongly verified. The intra-rater reliability was formally assessed using the methods outlined below. For the inter-rater analysis, each coder's assessments were merged to maximise the number of observable events, then the same methods were followed as per below.

The inter- and intra-rater analyses were based on the non-parametric approaches originally proposed by Bland and Altman[32] and later applied to sports performance analysis by Cooper et al.[83] The differences in the number of events identified for each assessment were calculated and the means, medians and Bland-Altman plots were inspected. Frequency tables were calculated after matching each event and are reported along with the proportion of total agreement in sequence and their 95% confidence interval,[83] ignoring differences in labelling between head impacts, prolonged contacts and probable head impacts.

Video verification

The video and sensors dataset are described using mean and standard deviation (SD), and median and interquartile range (IQR). The sensitivity and positive predictive value (PPV) to record events observed on video was calculated with their 95% confidence intervals (CI) for each sensor and each possible combination of sensors, and for every type of impact (Figure 4.7). The sensitivity of the sensors was determined as different if their 95% CI did not overlap.

The analyses then focused on head impacts and prolonged contacts to the head, as those were the only types of punches that were identified as scoring punches by a boxing judge (see below). The sensitivities and their 95% CI were plotted for each sensor and compared with respect to the type of head impact (direct vs. indirect), the impact location (in bins), the proximity of the impact to the sensor's location or to the face, and by participant.

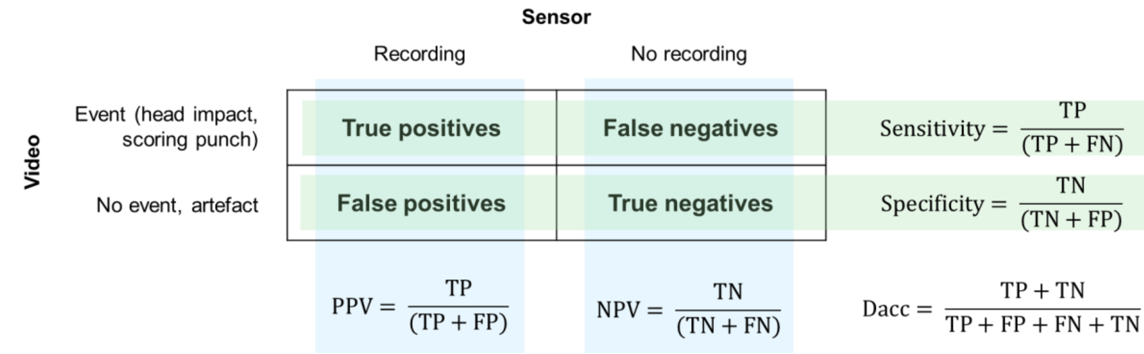


Figure 4.7. Sensor and video events classifications and formulas for the various metrics used in the study. TP: true positive, FN: false negative, FP: false positive, TN: true negative, PPV: positive predictive value, NPV: negative predictive value, Dacc: diagnostic accuracy.

Scoring punches

We enlisted an active AIBA-certified boxing judge and referee, to provide insights on whether the sensors were accurately capturing events that may be relevant from a scoring or refereeing perspective.[176] The certified judge reviewed 30 randomly selected 1-minute-long video segments of sparring (9% of the total duration of the video-verification dataset), including various boxers in both sparring areas and all available angles of video. She identified any punch sustained by a given participant that would influence her assessment of a boxer as a judge ("scoring punches")[176] and any hard punch that seemed to "shake" the boxer and that a referee would be looking for to assess whether the boxer is fit to continue. The term "scoring punch" hereby reflects an impact that landed effectively, i.e., with good technique.[387] Scoring punches are interesting to us because they contact cleanly with the head and may therefore transmit a higher force. The judge also identified any foul punches, which have a higher injury potential. Eight 1-minute video segments were reviewed twice, blindly, to determine the judge's reliability. Total agreement between test and re-test was reached when the same punches were identified in both assessments. The proportion of agreement was calculated and reported. After combining both assessments to maximise the number of scoring punches, those were matched to sensor recordings. The sensitivity, specificity, positive and negative predictive values (PPV and NPV) and diagnostic accuracy were calculated with their respective 95% CIs for each sensor and each combination of sensors. This analysis used scoring punches and non-scoring punches (i.e., punches that were coded by the primary coder but not identified by the judge as a scoring punch). The performance of the sensors was determined as different if their 95% CIs did not overlap.

Heavy blows

At the end of each sparring session, the participants answered questions about the perceived intensity of each bout they took part in and reported whether they had sustained "*one (or more) heavy blow to the head that "rocked" or "fazed" them, left them feeling dizzy, off-balanced or with blurry vision, even for a short time"*. Heavy blows were identified on video by the primary coder using the description reported by the participant (e.g., location on the ring, type of punch, single punch or part of a combination) and matched to sensor data. Heavy blows were qualitatively described.

4.5. Results

A total of 115 rounds of sparring were included for analysis. Sixty additional rounds were excluded for the following reasons: a patch stopped working (N = 24), video-related issues (e.g., recording not starting on time or participant out of view, N = 21), a patch fell off (N = 8); a patch reached capacity (N = 5), or rounds of body sparring (only body shots were allowed, N = 2).

Over the 115 rounds of sparring, 5168 video events were coded, 57% being definite head impacts or prolonged contacts to the head (Table 4.2). These two categories combined resulted in an average of 8.4 ± 5.1 contacts per minute of sparring (median: 7.7 [IQR: 5.1 – 10.3], range: 1.0 – 32.9)(see Figure 4.8 for an example round). The impacts to the front of the head totalled 69%, followed by the right (16%) and left (13%) sides of the head. Direct contact represented 30% of the head impacts and 11% of the prolonged contacts. Totals of 27%, 6%, and 1% of observed head impacts landed in the vicinity of the mouthguard, skin patch and headgear patch, respectively.

Totals of 1176 mouthguard, 3051 skin patch and 3215 headgear patch events were recorded over the duration of the study. Events occurring outside of any sparring rounds were automatically removed (6%, 31%, and 30% of the datasets for the mouthguards, skin and headgear patches, respectively)(Figure E.1, Figure E.2, Figure E.3, in Appendix E). After selecting the rounds to be included in the analysis, there were 695 mouthguard, 1578 skin patch, and 1690 headgear patch events to video-verify.

Table 4.2. Sensor events numbers and sensitivities for the impacts observed on video.

	Video events (%)	Mouthguard			Skin patch			Headgear patch		
		Rec+ (%)	Rec-	Sensitivity	Rec+ (%)	Rec-	Sensitivity	Rec+ (%)	Rec-	Sensitivity
All head impacts	3998 (77)	675 (97)	3323	16.9% (15.7% - 18.1%)	1519 (97)	2478	38.0% (36.5% - 39.5%)	1622 (97)	2375	40.6% (39.1% - 42.1%)
Definite + prolonged contact	2960 (57)	645 (93)	2315	21.8% (20.3% - 23.3%)	1399 (89)	1560	47.3% (45.5% - 49.1%)	1497 (89)	1462	50.6% (48.8% - 52.4%)
Definite head impact	2608 (50)	603 (87)	2005	23.12 (21.5% - 24.8%)	1264 (81)	1343	48.5% (46.6% - 50.4%)	1352 (80)	1255	51.9% (49.9% - 53.8%)
Prolonged contact	352 (7)	42 (6)	310	11.9% (8.8% - 15.8%)	135 (9)	217	38.3% (33.3% - 43.61)	145 (9)	207	41.2% (36.1% - 46.5%)
Probable head impact	1038 (20)	30 (4)	1008	2.9% (2.0% - 4.1%)	120 (8)	918	11.6% (9.7% - 13.7%)	125 (7)	913	12.0% (10.2% - 14.2%)
Clinch	286 (5)	16 (2)	270	5.6% (3.3% - 9.0%)	39 (2)	247	13.6% (10.0% - 18.2%)	48 (3)	238	16.8% (12.8% - 21.6%)
All body impacts	867 (17)	2 (0)	865	0.2% (0.0% - 0.9%)	10 (1)	857	1.1% (0.6% - 2.1%)	7 (0)	860	0.8% (0.4% - 1.7%)
Body impact	690 (13)	2 (0)	688	0.3% (0.0% - 1.1%)	8 (1)	681	1.2% (0.5% - 2.3%)	7 (0)	682	1.0% (0.4% - 2.1%)
Probable body impact	177 (3)	0 (0)	177	0.0% (0.0% - 2.0%)	2 (0)	175	1.1% (0.2% - 4.0%)	0 (0)	177	0.0% (0.0% - 2.0%)
Artefacts	16	0 (0)	16	0.0% (0.0% - 10.4%)	1 (0)	15	6.2% (0.3% - 29.1%)	3 (0)	13	18.7% (5.0% - 44.6%)
Sensor false positives		2 (0)			8 (1)			9 (1)		

Rec+: recording present, Rec-: recording absent.

Intra/inter-rater analyses

Over the ten 1-minute video segments of sparring action selected for intra/inter-rater analyses, the number of events recorded varied between 5 and 44. Within each rater, the mean differences between the test and re-test assessments were -1.1 ± 2.3 events per minute of sparring for the primary coder (i.e., more events identified during the re-test) and 1 ± 1.6 events for the boxing expert (i.e., fewer events during the re-test). The inter-rater results revealed a mean difference of 3.2 ± 2.7 events, such that the primary coder identified more events per minute than the boxing expert. For all intra- and inter-rater tests, the Bland-Altman plots showed that the number of differences increased with the number of events.

Overall, most head impacts (accepting differences in labelling between head impacts, prolonged contacts and probable head impacts) were matched between the test and retest for both coders and the proportion of total agreement in sequence reached 75.0% (95%CI: 65.9-82.3%) for the boxing expert, 90.2% (95%CI: 74.7-96.7%) for the primary coder, and 80.8% (95%CI: 54.8-93.6%) for the inter-rater analysis. The boxing expert missed a total of 12 obvious definite head impacts over the test-retest, half of which appearing of moderate intensity (Table 4.3A). The primary coder missed one obvious definite head impact, for which the video suggested an impact of light intensity (Table 4.3B). After merging each coder's two assessments, no definite head impacts were omitted by the primary coder when using the boxing expert's assessment as the reference (Table 4.3C). Relative to the direct/indirect head impact descriptor, the primary coder viewed 31% of the impacts as direct, against 73% for the boxing expert, resulting in a proportion of agreement of 56%.

Table 4.3. Frequency tables for A) the boxing expert intra-rater reliability, B) the primary coder intra-rater reliability, C) the boxing expert/primary coder inter-rater comparison. The light green cells represent the categories of head impacts, while the darker green cells highlight the diagonal of the table, i.e., the events that have been similarly identified and labelled for both test and retest assessments or between the two raters. The light red cells represent the head impacts that a coder identified for only one assessment, with the darker red cells highlighting the categories that we are most interested in (obvious definite head impacts and prolonged contacts missed during one of the assessments).

A) Boxing expert

Retest \ Test	Definite head impact	Prolonged contact	Probable head impact	Clinch	Body impact	Probable body impact	Missing: obvious	Missing: probable	Total
Definite head impact	68	1	2	1	2		5	1	80
Prolonged contact	1	1							2
Probable head impact	3	1	1			1	1	1	8
Clinch			1	2					3
Body impact					19	1	2	1	23
Probable body impact					2				2
Missing: obvious	7	1	2	2					12
Missing: probable	4		4			1			9
Total	83	4	10	5	23	3	8	3	139

B) Primary coder

Retest \ Test	Definite head impact	Prolonged contact	Probable head impact	Clinch	Body impact	Probable body impact	Missing: obvious	Missing: probable	Total
Definite head impact	63	6	5	4			1	1	80
Prolonged contact	3	6	2						11
Probable head impact	9	2	15	1	1		1	8	37
Clinch	1			9	1	1		1	13
Body impact					15	3	2		20
Probable body impact					6	2			8
Missing: obvious									0
Missing: probable			1		1	1			3
Total	76	14	23	14	24	7	4	10	172

C) Inter-rater

Boxing expert \ Primary coder	Definite head impact	Prolonged contact	Probable head impact	Clinch	Body impact	Probable body impact	Missing: obvious	Missing: probable	Total
Definite head impact	70	8	11						89
Prolonged contact	1	1		1					3
Probable head impact	5		9					1	15
Clinch				5					5
Body impact					18	5			23
Probable body impact						2		1	3
Missing: obvious	1	2		1	2				0
Missing: probable	3		18	4	1	2			28
Total	80	11	38	11	21	9	0	2	166

Video verification

We were able to match 693 of the 695 mouthguard events to a video event, 1569 of the 1578 skin-patch events to a video event and 1678 of the 1690 headgear-patch events to a video event, meaning the true positive recordings represented over 99.5% of each sensor dataset (Figure 4.8, Table 4.2). It was common for the participants to sustain several impacts within a couple of seconds, and therefore between 14 and 17% of the true positives could not be matched with one specific video event (they were marked as true positives, but the type and impact location were uncertain).

Overall, the patches were triggered twice more often than the mouthguard, resulting in higher sensitivities (Table 4.2). Specifically, definite head impacts triggered the sensors more frequently than other types of impacts, with sensitivities reaching 23.1%, 48.5%, and 51.9% for the mouthguard, skin patch, and headgear, respectively. The PPV for a sensor recording to be directly associated with a head impact (definite head impacts, prolonged contacts, and probable head impacts combined) was 97.1% for the mouthguard, 96.3% for the skin patch, and 96.1% for the headgear patch. Combining two or three sensors did not improve the sensitivity to head impacts or PPV when compared with the most sensitive individual sensor (Figure 4.9).

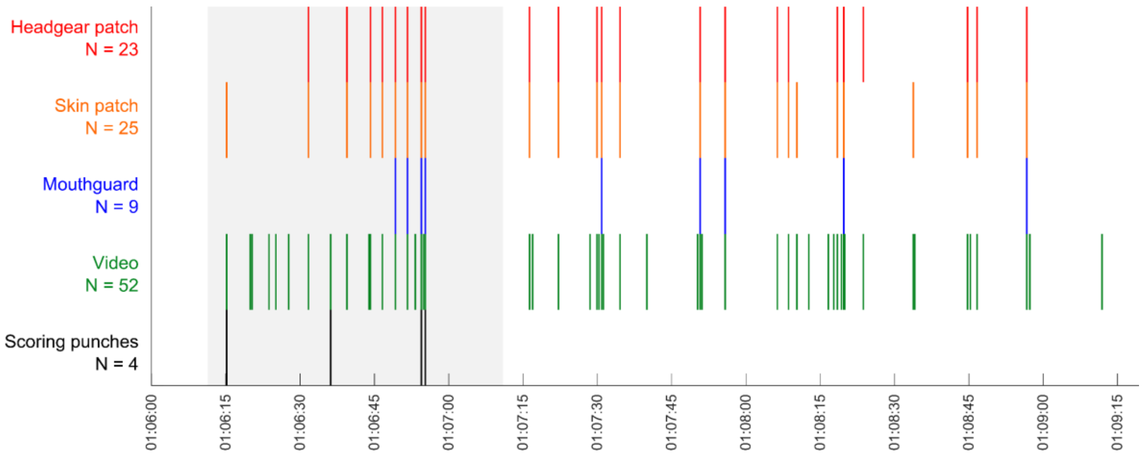


Figure 4.8. Representation of the events over time for one round of sparring (3 minutes), after video verification. The scoring punches were only coded for the first minute of the round, shaded in grey. Each vertical line represents one event and all lines are the same width; lines that appear thicker mean there were two or more events in close proximity. Video events include head impacts and prolonged contacts only.

Few false positive recordings were captured by the sensors: 2, 9, and 12 by the mouthguard, skin patch, and headgear patch, respectively. For the 12 corresponding events, four seemed to be associated with a body punch landing on the participant's forearms and the headgear being touched by the participant's own gloves, three while the participant was throwing a punch, three while he/she was adjusting his/her headgear, and the remaining two were undetermined. Out of the 16 instances where the video coder observed the participants adjusting their headgear, three triggered a sensor.

All three sensors were more sensitive to direct than indirect head impacts but similarly sensitive to impacts to the face area (Figure 4.9). For the mouthguard, the proximity of the punch to the sensor did not strongly affect its sensitivity (25% and 21%, for in and out of the vicinity respectively)(Figure 4.9), but some impact locations led to a sensitivity different from the overall sensitivity (Figure 4.10). Both patches were substantially more sensitive to impacts occurring in their vicinity (72 versus 45% of events recorded for the skin patch and 78 versus 50% for the headgear patch). The skin patch's sensitivity was increased for impacts to the right side of the head (where the patch was attached) and to the back. The headgear patch's sensitivity was increased for impacts to both sides of the head and even more for impacts to the back (where the patch was attached). The analysis of the sensitivity by participant suggested that there might be inter-participant differences, but that those were consistent across all three sensors (e.g., all three sensors seem more sensitive for participant 03 than for participant 04)(Figure 4.11).

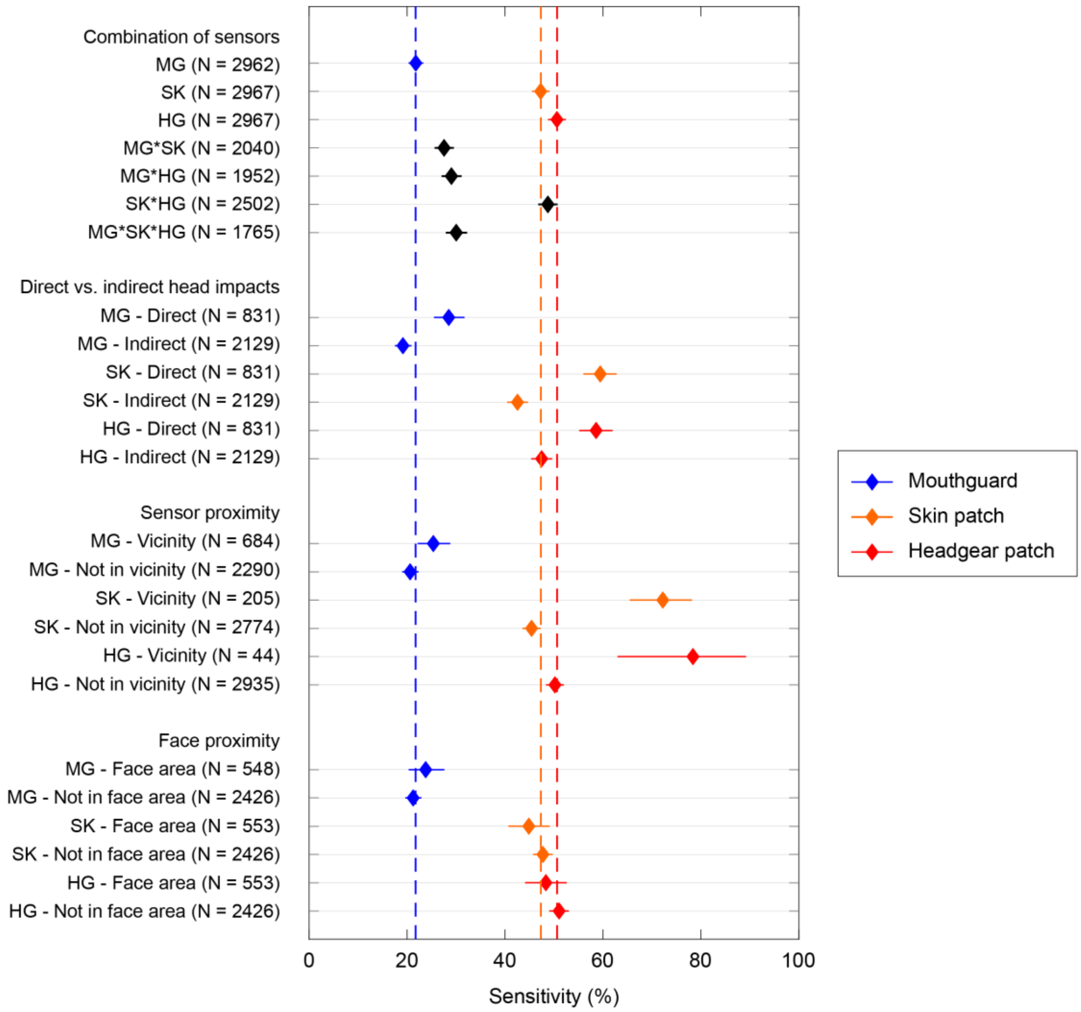


Figure 4.9. Sensitivity of the sensors to definite head impacts and prolonged head contacts relative to various parameters. The event numbers correspond to the video-coded events for each category. The diamonds represent the sensitivity and the lateral bars the 95% confidence intervals. The coloured dashed vertical lines represent the overall sensitivity for each sensor. MG: mouthguard, SK: skin patch, HG: headgear patch

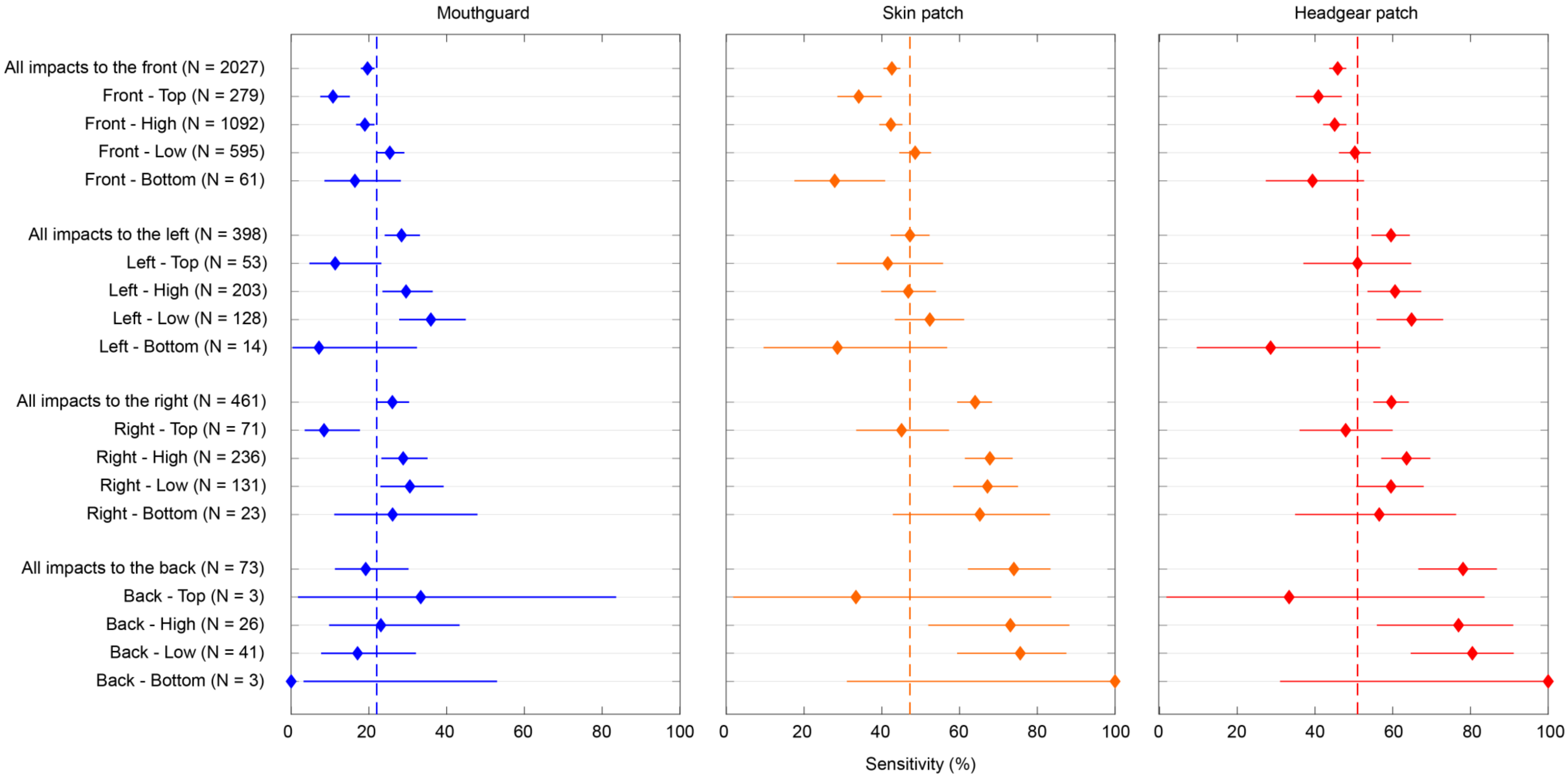


Figure 4.10. Sensitivity of the three sensors to head impacts and prolonged head contacts at each impact location bin (determined from the video events). The diamonds represent the sensitivity and the lateral bars the 95% confidence intervals, and the coloured dashed vertical line represent the overall sensitivity of the sensor.

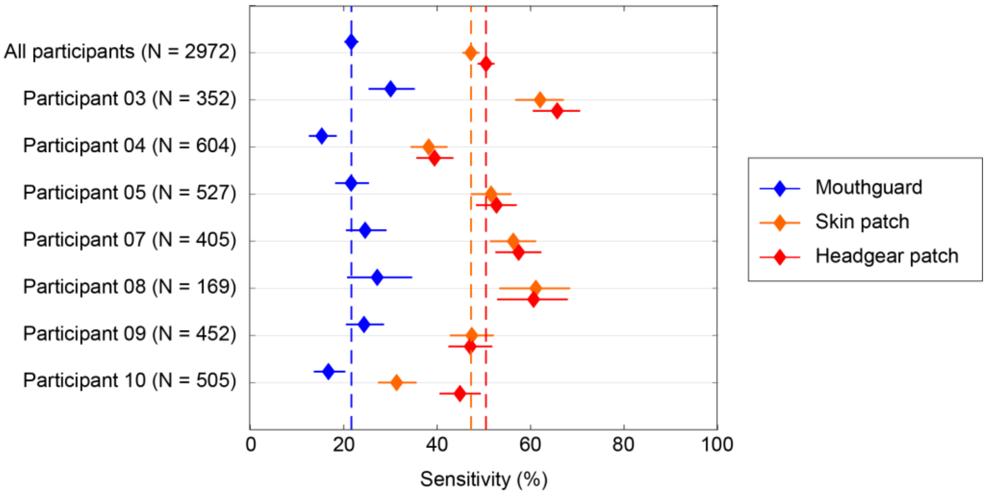


Figure 4.11. Sensitivity of all three sensors to head impacts and prolonged head contacts for each participant. The diamonds represent the sensitivity, the lateral bars the 95% confidence intervals, and the coloured dashed vertical lines the overall sensitivity.

Scoring punches

The number of differences between the certified judge's test and retest assessments showed a median of 0 differences per minute of sparring, ranging from -1 to 3. The assessments were in total agreement for 50% of the 1-minute video segments (95%CI: 21.5%-78.5%). When allowing for one difference per minute of sparring, the proportion of agreement reached 75.0% (95%CI: 40.9%-92.9%).

A total of 50 scoring punches were sustained by the participants over the 30-minute sample of sparring that was reviewed by the certified judge. All scoring punches were associated with a head impact (n = 44) or a prolonged contact to the head (n = 5), apart from one body punch. There was no foul punch or punch that visibly shook any of the participants in this sample. The number of scoring punches to the head averaged 1.6 ± 1.8 punches per minute of sparring per participant (range: 0 - 8, median: 1 [IQR: 0 - 2.25]). Mouthguard, skin patch or headgear patch recordings were matched to 17, 42 and 38 scoring punches to the head, respectively. This resulted in a sensitivity of 35%, 86% and 78% and a specificity of 90%, 76%, and 75% for the mouthguard, skin and headgear patch, respectively (Table 4.4). Any combination of sensors that included the mouthguard was most accurate overall (diagnostic accuracy between 84% (95%CI: 80 – 87%) and 89% (95%CI: 85 – 92%)), although not statistically more accurate than the mouthguard alone (the confidence intervals overlapped).

Heavy blows

Five heavy blows to the head were reported by three participants over the 18 athlete exposures. Two heavy blows were reported by the same participant on two different days and from two different sparring partners; both resulted from rear hooks to the left ear area, and all three sensors recorded an acceleration event in both cases. Two other heavy blows were from a jab-cross-jab-cross combination but could not be identified on video, and the last one was from an uppercut but was not collected due to video issues.

Table 4.4. Frequency tables and diagnostic measures (95% confidence intervals) for each individual sensor and all sensor combinations to correctly record scoring punches to the head. This table includes the sensors recordings where it was impossible to be sure to which event they were associated but were matched to the best of our understanding of what triggers the sensors (1 mouthguard, 3 skin patch and 3 headgear patches were potentially associated with a scoring punch).

			MG		SK		HG		MG*SK[a]		MG*HG[a]		SK*HG[a]		MG*SK*HG[b]	
		N	Rec+	Rec-	Rec+	Rec-	Rec+	Rec-	Rec+	Rec-	Rec+	Rec-	Rec+	Rec-	Rec+	Rec-
Video events[c]	Scoring punch	49	17	32	42	7	38	11	17	7	17	11	37	6	17	6
	No scoring punch[d]	387	37	350	94	293	97	290	31	287	31	284	76	272	29	268
Sensitivity			35% (30% - 39%)		86% (82% - 89%)		78% (73% - 81%)		71% (66% - 76%)		61% (55% - 66%)		86% (82% - 89%)		74% (69% - 79%)	
Specificity			90% (87% - 93%)		76% (71% - 80%)		75% (71% - 79%)		90% (86% - 93%)		90% (86% - 93%)		78% (74% - 82%)		90% (86% - 93%)	
PPV			31% (27% - 36%)		31% (27% - 35%)		28% (24% - 33%)		35% (30% - 41%)		35% (30% - 41%)		33% (28% - 38%)		37% (32% - 43%)	
NPV			92% (89% - 94%)		98% (96% - 99%)		96% (94% - 98%)		98% (95% - 99%)		96% (94% - 98%)		98% (96% - 99%)		98% (95% - 99%)	
Diagnostic accuracy			84% (80% - 87%)		77% (73% - 81%)		75% (71% - 79%)		89% (85% - 92%)		88% (84% - 91%)		79% (75% - 83%)		89% (85% - 92%)	

MG: mouthguards, SK: skin patch, HG: headgear patch, Rec+: recording present, Rec-: recording absent.

[a] Both sensors have a matched recording, or both did not record (i.e., the instances where only one out of the two sensors recorded an event were excluded).

[b] All 3 sensors have a matched recording, or all 3 sensors did not record anything (i.e., the instances where only one or two sensors recorded an event were excluded). All 3 sensors were in agreement for 320 video events (73% of this sample dataset).

[c] Video events include every type of events: head impacts, prolonged contacts and probable head impacts, clinches, definite and probable body impacts.

[d] The "No scoring punches" are all events that were observed on the video segments by the primary coder, but that were not identified by the certified judge as a scoring punch.

4.6. Discussion

The goals of this study were to: (1) compare the number of acceleration events recorded by a mouthguard, a skin patch and a headgear patch used simultaneously; (2) assess whether the type or location of impact, or the participant, affects the recording of an acceleration event; and (3) evaluate the accuracy of the sensors to record scoring punches or heavy blows. Overall, the patches recorded more than twice as many events as the mouthguard did. All three sensors recorded head impacts with very few false positives, leading to positive predictive values for head impacts over 96%. On the other hand, a large proportion of definite head impacts and prolonged contacts identified on video did not generate a sensor recording (78% for the mouthguard, 53% for the skin patch and 49% for the headgear patch). All three sensors were more likely to be triggered by definite head impacts, direct head impacts and impacts landing in the vicinity of the sensor's location. While this was true for all three sensors, the mouthguard seemed to show a more stable behaviour than the patches. With respect to scoring punches, the skin and headgear patches were more sensitive than the mouthguard (86 and 78% vs 35%) but less specific (76 and 75% vs 90%).

Intra/inter-rater analyses

The intra/inter-rater analyses aimed to determine if a non-boxing expert could accurately and reliably identify contacts to the head and the body. The primary coder identified more contact events and was more reliable than the boxing expert (90 % vs. 75% agreement). Because of our inclusive approach, this was a positive outcome as it may have minimised the number of events missed (23% of impacts were reported missing in the first round of video coding in another study,[136] vs. <1% in this study). Our findings suggest that the proficiency to identify head contact events on video does not rely on specialist knowledge of the sports actions. Rather, the primary coder had substantially more experience using the video software and had more vested interest in the study outcomes.[407] Sufficient time should be allocated for the coders to master the tools and appropriate and clear definitions of events should be used to minimise variability.

Numbers of acceleration events

We observed a total of 2960 definite head impacts and prolonged contacts in about 6 hours of active sparring, and have identified matches from the mouthguard, skin, and headgear patches for 22%, 47%, and 51% of them. From other studies in team sports, mouthguards and skin patches sensitivities averaged 71% with a 5-10 g triggering threshold (range: 41 to 98%).[67, 136, 220, 224, 339] In our study, the calculation of sensitivity was governed by our highly inclusive video coding approach, that does not match the methods used by the sensors, which are designed to ignore low-magnitude impacts. A less inclusive video coding approach may have resulted in higher sensitivity numbers. Given that the sensors and video coding approaches were not equally inclusive, the overall sensitivity cannot be used alone to evaluate the performance of the sensors. However, the differences in sensitivity can help understand how extrinsic and intrinsic factors affect the sensors' capacity to be triggered.

The proportion of false positive acceleration events recorded by the sensors was low (0.3-0.7%). In the conservative scenario where all acceleration events associated with body impacts and clinches were artefacts, this proportion was still lower than 4%. These are amongst the lowest numbers reported, as false positive proportions (FP/(FP+TP), or 1-PPV) have been shown to vary between 3.6 and 89% with instrumented mouthguards in rugby,[198] soccer[339] and American football[136, 198, 220] and 15 to 90% with skin patches in Australian rules football[258] and soccer.[322] The difference may have arisen from our definition of true/false positives and our highly inclusive video criterion. A more restrictive criterion, e.g., classifying as true positives only the impacts that visibly modify the trajectory of the head,[136] would likely have resulted in a higher proportion of false positives.

The low proportion of false positive sensor events may also be a consequence of the sport we studied. First, there is no running, and only limited jumping during boxing, two common causes of false positives from patches.[322, 339] Second, a round of sparring means three minutes of intense activity, during which there is barely any time to take the mouthguard or headgear off or do anything that might trigger spurious events. Third, the sparring was dictated by a bell announcing the start and end time of each round, i.e., we rarely had one participant rest while all others were sparring, as is often the case in team sports. It was then particularly easy to use the videos and the sound of the bell to automatically exclude all acceleration events that occurred outside of sparring time, for all participants. This step was essential as it excluded up to 31% of a sensor' raw dataset, containing only spurious events.

Effects of factors on the proportion of events being triggered

Sensor

Both patches were triggered more than twice as often as the mouthguard, despite all sensors being programmed to be triggered using a 10-g threshold. The only studies comparing mouthguard, skin patch and headgear-based devices were conducted in the laboratory with semi-controlled impacts,[219, 228, 418] making a comparison with our study difficult. Yet, these studies and others have shown that skin patches, skull caps and helmets are not rigidly coupled to the skull.[184, 418] We observed the motion of the headgear in several instances during the video coding step:

the punch would land on the headgear in one frame, the headgear would visibly slide back in the next frame, followed by the motion of the head one frame later. Sensors that are not well coupled to the skull measure a different motion than that of the skull upon impact and generally overestimate the peak linear acceleration.[184, 219, 367, 418] Such poorly-coupled sensors will therefore more often record accelerations above the triggering threshold than well-coupled sensors, resulting in higher numbers of impacts, which may explain why the patches recorded twice as many events than the mouthguard. Nevertheless, we cannot exclude the possibility that differences in design or technology between the sensors may also have contributed to this disparity.

Type of impact

All three sensors were predominantly triggered by definite head impacts and prolonged contacts, representing more than 90% of each sensor's dataset. While the patches were twice as sensitive to definite head impacts as the mouthguard, they were three-to-four times more sensitive to prolonged contacts and probable head impacts. These types of impact may therefore lead to the skin and headgear patches moving sufficiently to trigger the sensors, while the acceleration measured by the mouthguard is more frequently below the threshold. The sensors were rarely triggered by body impacts (0-1% of the sensors datasets, compared to 17% for the video dataset), which also showed a light contact to the head or headgear in most cases. This small proportion suggests that a body punch does not generate head motion high enough for the sensors to be triggered, probably due to the relatively low momentum generated by a punch. All sensors were more often triggered by direct than indirect impacts, which could be a result of direct impacts being of higher magnitude; this hypothesis will be tested in the next chapter. However, because of the low agreement between the boxing expert and the primary coder on the application of the direct/indirect impact definition, this result must be interpreted with caution.

Proximity

Our findings highlighted an increased sensitivity to impacts in the vicinity of the sensor (from 21 to 25% for the mouthguard, 45 to 72% for the skin patch, and 50 to 78% for the headgear patch). If the sensors were rigidly fastened to the skull, the location of the impact should not influence their capacity to be triggered to that extent. If the sensors were not well coupled when hit directly, they would move at a faster rate than the skull, leading to higher accelerations and a new recording. We observed the dislocation of the headgear and noted that the skin patch was often loose due to intense sweating (and in cases may have been kept in place by the headgear), which show poor coupling. Another study excluded direct impacts to the instrumented mouthguard, as they "*may have produced a sharp spike in the acceleration data",*[389] which could explain why a sensor would be triggered more easily when the punch lands in its proximity. This raises an important point: while most of the sensor recordings were video-verified, it was impossible to know if a recording was triggered by true head motion or by skull/sensor decoupling. An examination of the kinematic traces will be conducted in the next chapter to better understand the effects of impact location.

Participant

Finally, we observed differences between participants in terms of sensor sensitivity to impacts. These differences may originate from boxing styles and our inclusive approach. For example, participant 10 mostly kept his guard up (i.e., his own gloves against his forehead to protect himself), leading to numerous light-looking impacts to his gloves, and participant 04 admitted purposefully protecting herself less when she "*knew [she] could take it*". These behaviours could be reflected by a higher number of impacts that were observed on video but below the triggering threshold. Contrarily, participant 03 tended to avoid the punches, reducing the number of events altogether.

Accuracy to record scoring punches and heavy blows

The identification of scoring punches by the AIBA-certified judge allowed us to determine the accuracy of the sensors to record impacts that landed effectively on the head. The PPV was similar for all three sensors (28-31%), suggesting they are as useful at differentiating a scoring from a non-scoring punch, as long as an event was recorded. However, the sensitivity of the patches was much higher than the mouthguard's (78-86% versus 35%), meaning the chance they will record a scoring punch in the first place is higher. It is concerning that impacts of interest, such as the scoring punches or injurious impacts in other studies,[199] can be missed by the mouthguard. To summarize, if we give priority to scoring punches over any other impact, then sensitivity takes priority over specificity or diagnostic accuracy, and the patches are therefore more useful than the mouthguard to quantify the number of clean punches, which may transmit a higher amount of force. However, because of their moderate specificity, it would not be recommended to use the patches on their own to score fights or determine an athlete's impact load.

Implications

Both patches recorded larger numbers of acceleration events than the mouthguard, with a high proportion of true positives, and were more sensitive to scoring punches. While both patches were sensitive to impact location, the occurrence of impacts to the back of the head was lower than to the side (2% vs. 16%), reducing the issue of an over-powered proportion of events recorded to the patch side of the head. The headgear patch is also easier to use: it adheres better than a skin patch and is easier to check or change without the boxer having to remove his/her headgear. Therefore, to use as an impact counter, i.e., to measure how many times a boxer's head is made contact with, independently of magnitude, a headgear patch is the most appropriate solution out of the sensor configurations we tested.

However, our analyses do not allow us to conclude whether the sensors can inform injury risks. Without knowing if an acceleration event is an artefact due to skull/sensor decoupling or a valid measurement of sudden head motion, the number of events recorded by a sensor may be overestimating the exposure. Our assessment of the Prevent Biometrics and CSx classification algorithms suggests that some acceleration events present kinematic characteristics that lead the algorithms to misclassify them as spurious (Appendix F). The analysis of the kinematic signals in Chapter 5 will shed light on the prevalence of artefacts and is needed to determine if the sensors can be used to better understand exposure and injury risks.

Strengths and limitations of the study

Boxing provided a suitable environment for the verification of head impacts. The risks of a participant being out of view of all cameras were limited, as were the risks of obstruction, and the camera setup allowed us to observe discrete events, in opposition to contact periods where several impacts occur in a short time frame.[67, 220] Furthermore, amateur boxing does not allow extended clinching or wrestling, making the contacts to the head more easily identifiable. As a result, less than 0.2% of sensor events could not be verified, which differs substantially from the 15-20% proportion of unverifiable sensor events reported in previous studies.[5, 410] Nonetheless, the number of probable head impacts, i.e., when contact to the head could not be asserted, represented 20% of all our video events. It is possible that additional cameras or better positioning may have reduced this proportion, but we were restricted by the number of cameras available, the need to capture two sparring areas at the same time, and the space available in the gym.

As stated above, our highly inclusive approach to identifying contact events on video is an important limitation of the study, as it did not match the sensor's processes and included minimal-magnitude events that are likely of little interest in the investigation of concussive injury risks. Our approach likely underestimated the sensors' sensitivity, but also over-estimated the PPV, as it was common for the skin and/or the headgear patches to record an event while the punch appeared to be of little effect on the boxer's head motion. We did not attempt to visually estimate the severity of each head impact from the video, as has been done previously.[136, 220] The reliability of such an approach relies on clear definitions and criteria, which we did not find satisfactory enough. A related limitation is that a single coder identified head impacts for the whole dataset, while multiple raters may be preferred to establish reference values.

It is important to note that the mission of a boxing judge is a complex one and that identifying scoring punches is not straightforward.[66, 176] As a result, the certified judge in this study showed moderate intra-rater reliability. While the judge was reviewing the video at full speed and noting timestamps of scoring punches, the identification of scoring punches may have been more reliable and accurate if we had shown and asked the judge to categorise each individual impact.

4.7. Conclusion

In summary, this study investigated the capability of multiple head impact sensors to record video-observed head impact events. About three-fourths of video events were missed by the instrumented mouthguard, and half by the skin and headgear patches, but we hypothesise that many missed events were below the acceleration threshold to trigger the sensors and are of little interest relative to injury risks. The number of spurious events was one of the lowest ever reported in head impact research and as a result, the positive predictive value was high for all sensors. Although the mouthguard was the least affected, all sensors were more sensitive to impacts occurring in their proximity, suggesting that all sensors may present varying degrees of decoupling from the skull. Because of its ease of use and its higher sensitivity to scoring punches, the headgear patch could be used as an impact counter. However, investigation of the kinematics is needed to advise on the sensors' potential for measuring exposure and informing injury risks.

Chapter 5 - Head impact kinematics during boxing sparring: assessment and comparison of an instrumented mouthguard, a skin patch, and a headgear patch

This manuscript is currently in preparation for journal submission.

Supplementary materials can be found in Appendix E.

5.1. Prelude

Chapter 4 has shown that the performance of the instrumented mouthguard, the skin patch, and the headgear patch to record head impacts observed on video was affected by the location of the impact. This suggested that the kinematic signals present different characteristics whether they result from an impact in close proximity to the sensor or not. Because skull/sensor coupling issues have been previously reported in the literature, the validity of the kinematic measurements from video-verified head impacts is explored in this chapter. Additionally, we aimed to compare the peak accelerations measured by the three sensors for simultaneous events, using the mouthguard data as the reference.

5.2. Abstract

Background: While head impact sensors are deemed of acceptable validity when tested in the laboratory, several limitations related to the kinematics measured *in-vivo* have been highlighted.

Objectives: (1) Assess the quality of the raw kinematic traces recorded by three different head impact sensors; (2) determine the effects of signal quality on peak accelerations; (3) compare the peak accelerations for synchronous events across sensors.

Methods: Head impacts were collected with an instrumented mouthguard, a skin patch, and a headgear patch during amateur boxing sparring. The raw kinematic traces for 442 video-verified events were assessed for each sensor against a set of defined criteria and classified as "good", "questionable" or "bad". The proportion and 95% confidence interval of 'good' recordings were calculated for participants, type of impact and impact location. Non-parametric tests were used to relate signal quality and peak kinematics, and assess the association between synchronous 'good' recordings across sensors.

Results: The quality criteria were met for 53, 20, and 26% of events for the mouthguard, skin patch and headgear patch, respectively. 'Bad' recordings were more frequent for impacts occurring close to the sensor and showed significantly higher peak kinematics than "good"

recordings ($p < 0.001$). There was little to no association between the sensors' measurements for simultaneous recordings (Spearman's $\rho \geq 0.043$).

Discussion: The kinematic data, particularly from the patches, often reflected the motion of the sensor itself rather than the motion of the head.

Conclusions: The kinematics measured by the skin and headgear patches should not be used for the analysis of boxing-related exposure to head impacts as their magnitude is affected by the patches' poor coupling to the skull. Research is required to improve the coupling of the mouthguard to the skull and determine how to use exposure datasets where large proportions of data are missing because of measurement issues.

5.3. Introduction

Head impacts sustained during sports participation have been studied to better understand concussive injury risks. Head impacts have been described primarily using summaries of peak linear and angular accelerations (PLA and PAA, respectively), measured using sensors embedded in helmets,[115] mouthguards,[19] skin patches,[257] or placed in headbands.[56] The devices typically consist of a triaxial linear accelerometer and a triaxial angular rate sensor and are normally tested during controlled laboratory impacts to verify that they measure what a reference sensor measures to an acceptable level of accuracy.[93, 180, 232, 369] While these laboratory tests have some resemblance to real-life head impacts, the biofidelity of the interface between the anthropomorphic test device and the sensor, e.g., the device's scalp or mandible, is limited.[368, 393] Additionally, the type and range of impacts replicable in the laboratory does not reflect the variability observed on the field.

When evaluated in biofidelic environments (live humans in the laboratory or post-mortem subjects), various types of sensors have demonstrated motion relative to the skull, which led to errors in the measured kinematics.[12, 367, 418] For instance, the scalp can experience high dynamic deformations, resulting in the sliding of the scalp's outer layer (i.e., the skin) relative to the skull, which can be described by a low-stiffness elastic behaviour.[12] Due to the hyper-elastic properties of the scalp and the combined inertia of the skin-based sensor and underlying soft tissue, the sensor's motion lags and overestimates the skull's motion.[367, 418] Moreover, when using headgear-based devices, such as helmets, headbands or skull caps, the presence of hair and sweat,[4, 393] as well as the fit of the headgear on the head,[180] affect the mechanical properties of the interface between the skull and the sensor. This in turn modifies the relative motion of the headgear on the head and thus the kinematics measured by the sensor.[180, 184, 394]

Skin patches have been attached to the skin of cadaver heads subjected to free drops and shown to overestimate PLA by 64 ± 41% and PAA by 370 ± 456% when compared to reference sensors rigidly fixed to the skull.[367] Additional research was conducted with human participants performing soccer ball headings in a laboratory environment.[219, 418] Wu et al. demonstrated various levels of skull/sensor coupling for an instrumented mouthguard, a skin patch, and a sensor mounted in a skull cap, using high-speed video with ear-canal markers used as reference.[418] The authors measured relative motion between the ear canal and the mouthguard of less than 1 mm, while the skin patch displaced by up to 4 mm and the skull cap sensor by up to 13 mm. When comparing the skin patch and skull cap sensor measurements to the instrumented mouthguard (used as the reference for 3-dimensional kinematics), there were over-estimations of the peak accelerations, with averages ranging between +161% for the skin patch for PLA and +1433% for the skull-cap device for PAA.[418] Kuo et al. found, when following a comparable impact protocol, similar results for skin patch data, with errors over 100% for all peak kinematics compared to mouthguard data.[219]

The information available on the validity of the head impact sensors in true *in-vivo* settings, i.e., during sports participation, is scarce. There have only been two studies comparing devices during sports participation, both involving the Head Impact Telemetry (HIT) System, a helmet-based device, and the skin-based sensor xPatch, in American football.[93, 227] Cummiskey et al. mentioned differences in the distribution of impact severity between the two systems,[93] while Lennon reported weak but significant associations (R^2 = 0.19 for PLA, 0.02 for PAA), with significantly higher PAA measured by the xPatch relative to the HIT System.[227] Neither study applied a reference sensor that these measurements could be compared to. Other studies comparing sensors and using a mouthguard as the reference were performed in a laboratory setting, rather than *in-vivo*, and were limited by a small sample size (1 or 4 participants), a small number of impacts conducted (< 35 impacts), the magnitude of the head impacts (~10 g), the particularity of having only one type and direction of impacts being tested (soccer heading), and specific instructions for the participants to clench their teeth.[219, 418]

Moreover, previous research has identified issues in the sensors' raw kinematic traces. Mouthguards have been shown to measure spikes in acceleration when impacted directly while being worn.[389] This may explain the increased proportion of recordings classified as spurious by the Prevent Biometrics algorithm for impacts in the vicinity of the mouthguard in our sparring data (Appendix F, page 247). For skin patches, 86% of all acceleration events recorded during combat sparring were classified as bad, i.e., contained short duration spikes, rapid signal inversions or clipped signals.[343] These issues reinforce the idea that sensors can decouple from the head and measure a motion that is not representative of head motion.

The over-estimates in magnitude observed with loosely coupled devices are problematic when trying to study head impact exposure and risk of concussion. To help understand the validity and limitations of head impact devices, the present study compares the kinematics of video-verified head impacts measured simultaneously by an instrumented mouthguard, a skin patch, and a headgear patch, using an *in-vivo* observational cohort study design.

The study's aims were to:

1. Assess the quality of the raw kinematic traces for each sensor, by determining the types of issues present and their prevalence;
2. Determine the association between signal quality and peak kinematics; and
3. Compare the peak accelerations measured by the mouthguard, the skin patch, and the headgear patch, using only recordings of acceptable quality.

Mouthguards have been shown to provide a more adequate coupling with the skull than skin- and headgear-based sensors,[219, 418] therefore, the measurements from the instrumented mouthguard were used as the reference.

5.4. Methods

Overview

The data were collected following the methods described fully in Chapter 4. Seven competitive boxers were equipped during their weekly sparring session with a Prevent Biometrics instrumented mouthguard, a CSx patch taped to the right mastoid process, and a second CSx patch taped to the back of the boxer's headgear. The sparring sessions were filmed using five cameras placed around two sparring areas. In the previous chapter, we selected 115 3-minute-long rounds of sparring for which all three sensors collected acceleration events. In this chapter, we consider acceleration events that have been video-verified and that have time-matched recordings for all three sensors.

Instrumentation

The Prevent Biometrics Hybrid mouthguards (Prevent Biometrics Inc., Edina, MN) are boil-and-bite mouthpieces that incorporate a triaxial linear accelerometer (ADXL372, Analog Devices, Boston MA, range ± 200 g) and a triaxial angular rate sensor (BMG250, Bosch, Gerlingen Germany, range ± 35 rad.s^{-1}), both sampling at 3200 Hz. The sensors are located in the vicinity of the first left lateral incisor.[232] During use, any time the linear acceleration signal reaches 10 g on any axis, the system records a 50 ms acceleration event (10 ms pre- and 40 ms post-trigger) and the recording is transmitted wirelessly to the Prevent Biometrics iOS app. Prevent Biometrics provided us with the raw 3-degrees of freedom (3DOF) time-series data from the accelerometer and angular rate sensor for the purpose of the study. We also downloaded the fully-processed traces for linear acceleration (LA), angular velocity (AV) and angular acceleration (AA), which are normally available to the end-user. Communication with the manufacturer revealed that raw signals were filtered with a fourth-order Butterworth filter with a cut-off frequency of 200 Hz, with an additional filtering step being applied depending on data quality, which is determined by the algorithm (no additional details were provided on this additional step). The filtered data were then transformed to the head's centre of gravity (CG) based on a 50th percentile male NOCSAE headform's dimensions. A previous publication suggested that individual sensor outputs were also calibrated against a reference system,[16] but we did not obtain more information.

While there has been no published report of the validity of the Prevent Biometrics Hybrid mouthguard, two recent independent studies compared the company's custom-made and original boil-and-bite versions (a bulkier and stiffer version than the Hybrid).[198, 232] For helmeted American football-like impacts, the PLA, PAV and PAA measured by the Prevent Biometrics mouthguards were well correlated to the reference sensor's signals ($R^2 > 0.89$ and linear fit slopes between 0.8 and 1 for impacts ranging from 20 to 100 g and from 15 to 50 rad.s^{-1}). The mean relative error between the mouthguards and the reference were below 5% with one exception (the mean relative error for PAA measured by the boil-and-bite version was 8.3%).[232] The second study reported general high agreement between the Prevent Biometrics mouthguards and the reference system for unhelmeted impacts, with concordance correlation coefficients of 0.95 for the boil-and-bite, and 0.91-0.97 for the custom mouthguard.[198]

The CSx patches (CSx Systems Ltd, Auckland, New Zealand) measure 25 x 21 x 6 mm and weigh 4.5 g. The patches house a triaxial linear accelerometer (ADXL375, Analog Devices, Norwood, MA, linear range ±200 g, sampling frequency 3,200 Hz) and a triaxial angular rate sensor (ITG-3701, Invensense, TDK, San Jose, CA; angular range ±4000 °.s^{-1}, sampling frequency 8,000 Hz). A 50 ms-long event (5 ms pre- and 45 ms post-trigger) was recorded every time any linear acceleration axis measured a value over 10 g (i.e., the same threshold was used for all sensors in this study). The raw 3DOF linear acceleration and angular velocity signals were available via direct download at the end of each session.

Both the Prevent Biometrics and CSx systems involve a proprietary classification algorithm that automatically assesses whether the acceleration event is deemed a true head impact or a spurious recording. We obtained and included all events independent of the algorithms' classification, which performance was evaluated in Appendix F. The mouthguards also have an infrared proximity sensor to estimate how close the mouthguard is to the teeth. Prevent Biometrics provided us with the proximity sensor readings, which were also evaluated in Appendix F (Figure F.3).

The CSx patches raw data were processed using custom MatLab scripts (R2019a, Mathworks, MathWorks, Natick, MA). The time series were first checked for missing data: if more than 1 ms of consecutive data points were missing, the recording was excluded from further analysis (13 recordings each for the skin and headgear patches over the duration of the study). Signals containing gaps shorter than 1 ms were automatically interpolated using the MatLab *interp1* function with the '*spline'* option. The time series were then filtered using a fourth-order low pass Butterworth filter with a cut-off frequency of 200 Hz for both LA and AV. The AV signal was down-sampled to match the LA sampling frequency (from 8000 to 3200 Hz), and angular acceleration was calculated from the filtered AV signal using the 5-point stencil differentiation method. The axes of the sensors were aligned with the head's coordinate system (X-front, Y-right, Z-bottom, see Figure 5.1). The patches' location and orientation relative to the head's coordinate system were estimated for each participant using photographs and measurements of the participants' heads with and without headgear (described in the section *Estimation of the position and orientation of the patches relative to the head's centre of gravity*). These estimates were used to transform the linear acceleration to the head's CG using rigid body kinematics (Equation 5.1).

$$a_{CSx/CG} = a_{CSx} + \alpha_{CSx} \times r_{CSx/CG} + \omega_{CSx} \times (\omega_{CSx} \times r_{CSx/CG}) \qquad \text{Equation 5.1}$$

$a_{CSx/CG}$ is the LA calculated at the head's CG, a_{CSx} is the LA measured by the patch at its own location, α_{CSx} is the AA differentiated from ω_{CSx}, which is the AV signal measured by the patch, and $r_{CSx/CG}$ is the vector from the sensor's location to the head's CG.

To the best of our knowledge, there has been no assessment of the validity of the CSx patches. However, we have investigated the validity of the CSx mouthguards, which house the same electronics and use the same algorithms as the patches. The CSx mouthguards were on par with other head impact sensors but were limited by moderate validity and accuracy, especially for the PLA (see Chapter 3). Laboratory validation of the CSx patch was originally planned as part of this doctoral work, but the COVID-19 pandemic prevented travel to the intended laboratory (Vancouver, Canada), and the equipment needed was not available locally.

Estimation of the position and orientation of the patches relative to the head's centre of gravity

A series of anthropometric measurements were taken from the participants upon recruitment. Head length was measured as the maximum length of the head between the glabella and the occiput, and head breadth as the maximum horizontal breadth of the head above the attachment of the ears. The maximum headgear length and breadth were also measured. Profile and back photographs were taken with an iPad at the beginning of each sparring session, with and without headgear, displaying the patches attached to the head and headgear. The following landmarks were identified from the profile photographs and used as follows (Figure 5.1, A and B):

1. The porion (the superior edge of the external auditory meatus) and the inferior border of the orbital rim were used to define the X-Y plane of the head (the Frankfort plane).
2. The glabella and the occiput were used to calibrate head length (pixels to cm and vice-versa) from the measurements taken on the participants, and the front and back points of the headgear were used to calibrate headgear length.
3. The position of the CG along the X-axis (CG-X) was calculated 0.83 cm anterior of the porion.[29]
4. The Y-Z plane was calculated normal to the X-Y plane at the CG-X point.
5. The position of the CG along the Z-axis (CG-Z) was calculated 3.13 cm above CG-X.[29]

From the back photograph, the left and right points for the maximum horizontal breadth of the head and headgear were identified to define the Y-axis and the distance was calibrated from the head/headgear breadth measurement (Figure 5.1, B and D). The position of the CG along the Y-axis (CG-Y) was defined as the mid-point between the left and right sides.[29]

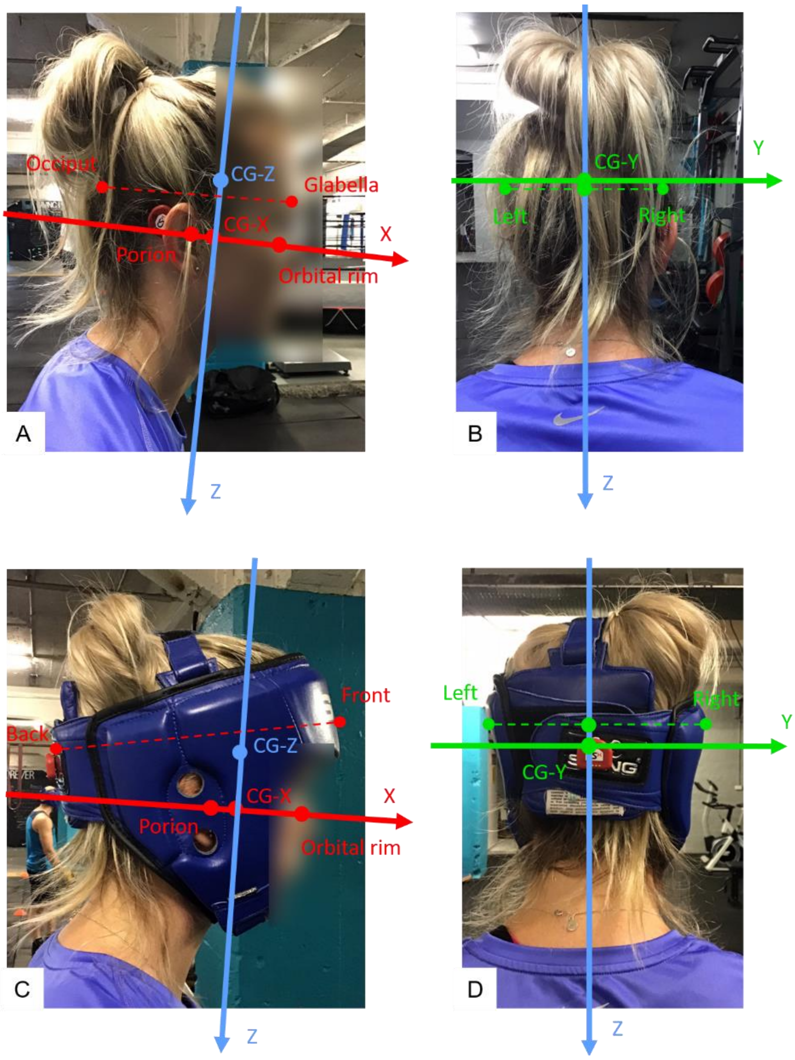


Figure 5.1. Identification of the head's coordinate system from anatomical landmarks. The solid lines represent the head's axes, and the dashed lines represent the measurements used to convert real distances from the anthropometric measures to pixels on the images, and vice-versa. The photographs of the bare head (A, B) were used to estimate the position and orientation of the skin patch and the photographs of the participant with headgear (C, D) were used for the headgear patch.

Once the origin (the CG) and the axes of motion were defined from the anatomical landmarks, the position vector and the Euler angles determining the orientation of the patches were estimated from the photographs. This was calculated for all photographs available and of sufficient quality (between one and four per participant) and was averaged to generate participant-specific parameters for the transformation to the head's CG. Anthropometric measurements, location vectors and Euler angles for the alignment of the patches to the head's coordinate system are presented in Table 5.2 for each participant. Custom rotation matrices were calculated for each participant and applied to the kinematic signals using the equations below.

$$R_{x''} = \begin{bmatrix} 1 & 0 & 0 \\ 0 & \cos(\psi) & \sin(\psi) \\ 0 & -\sin(\psi) & \cos(\psi) \end{bmatrix}$$

$$R_{y'} = \begin{bmatrix} \cos(\phi) & 0 & -\sin(\phi) \\ 0 & 1 & 0 \\ \sin(\phi) & 0 & \cos(\phi) \end{bmatrix}$$

$$R_z = \begin{bmatrix} \cos(\theta) & \sin(\theta) & 0 \\ -\sin(\theta) & \cos(\theta) & 0 \\ 0 & 0 & 1 \end{bmatrix}$$

$$R = R_z R_{y'} R_{x''}$$

Selection of acceleration events

The previous chapter identified 693, 1569, and 1678 mouthguard, skin, and headgear patch video-verified acceleration events occurring over 115 rounds of sparring. We excluded all acceleration events that could not be matched with certainty to a specific impact observed on video because of the high number of impacts observed within 1-2 seconds (116, 217, and 263 events excluded for the mouthguard, skin patch, and headgear patch, respectively). Next, we excluded the events that did not have a recording from all three sensors (115, 900, and 960 events excluded). These two steps ensured that the recordings across the three sensors corresponded to the same event. Finally, we omitted the events that were not associated with a "definite head impact" or a "prolonged contact" (as defined in Chapter 4, page 56), as these were the types of impacts that led to scoring punches (20 events excluded). A total of 442 head impact events with a matched recording for all three sensors were available for analysis.

Raw signals quality assessment

A preliminary visual assessment of the raw signals showed that a non-negligible proportion of recordings presented issues that were evocative of the decoupling between the sensor and the skull, similar to prior findings.[343, 389] We wished to focus the comparison of the kinematics across the sensors to recordings that appeared to be representative of the motion of the head, rather than of the independent motion of the sensor. To appraise the quality of the raw 3DOF LA and AV traces, criteria were defined from experience of previous laboratory and fieldwork and through discussions among the research team (Table 5.1, Figure 5.2, Figure 5.3). Each mouthguard and patches recording was manually classified by a single rater as satisfying the quality criteria ('Good'), 'Questionable', or not satisfying the quality criteria ('Bad').

Table 5.1. Definition of the quality classes and description of the criteria used to assess the raw signals.

Class	Definition
'Good'	Recordings for which all the traces look like what could be expected of an impact to the head measured by a rigidly-mounted sensor.
'Questionable'	The recording does not clearly meet any 'Bad' criteria, but confidence that the recording classifies as 'Good' is low.
'Bad'	One or several traces are evocative of the motion of a low-inertia object (i.e., the sensor itself) in opposition to a high-inertia object (i.e., the head). This was reflected by the presence of at least one of the following criteria: - rapid signal inversion (a full sinusoid within a few milliseconds) - short duration spike - peaks in linear acceleration and angular velocity being aligned in time - smooth motion suddenly interrupted - noisy signal suddenly becoming smooth - signal reaching sensor saturation - high-frequency and high-magnitude noise - flat line/missing data.

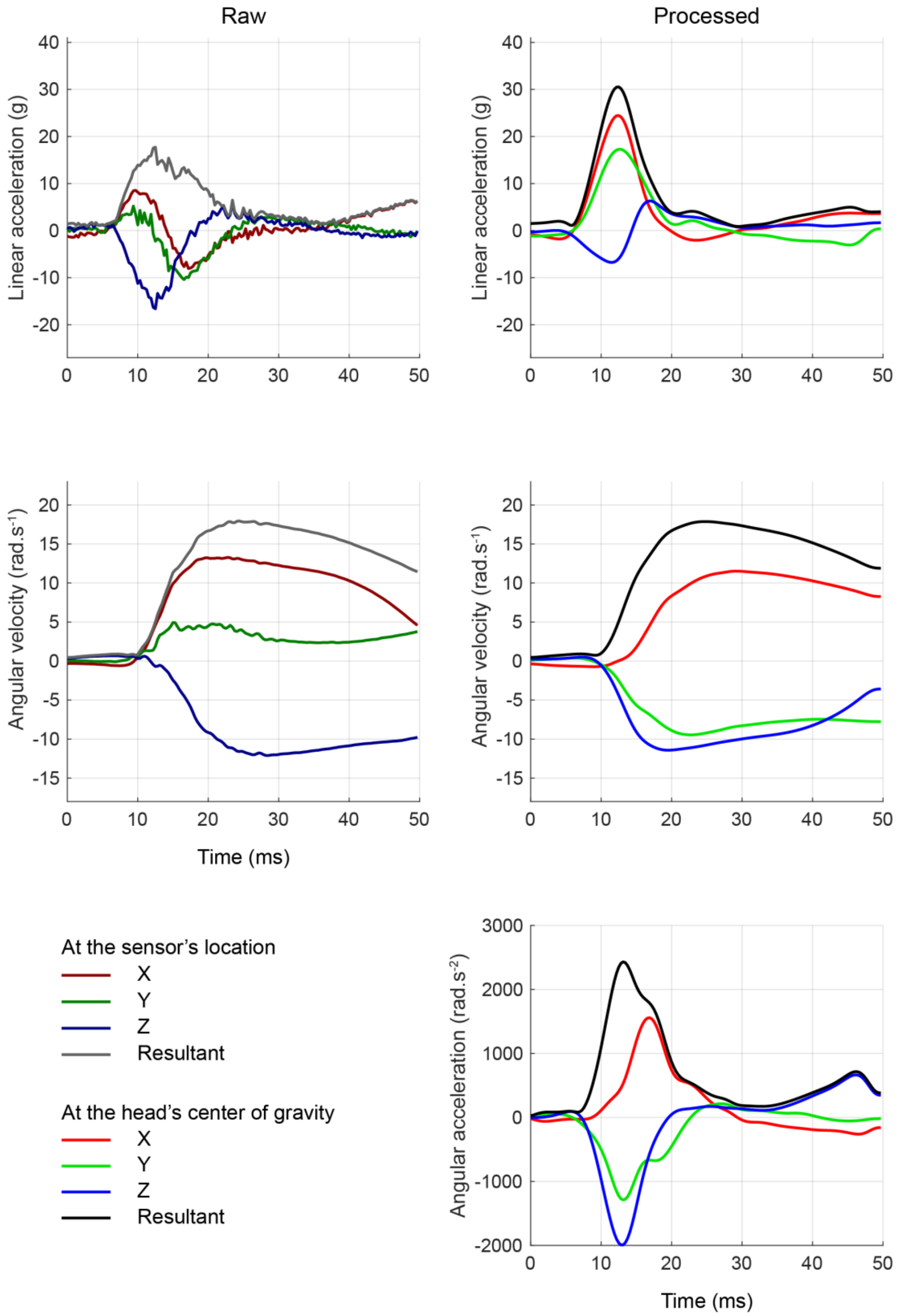


Figure 5.2. Example of a 'good' signal from the mouthguard, evocative of the motion of a high-mass, high-inertia object. The raw linear acceleration (top-left) and raw angular velocity (middle-left) were provided by Prevent Biometrics and are plotted at the sensor's location. Note the orientation of the sensor's axes is unknown for the raw data. The processed linear acceleration (top-right), angular velocity (middle-right) and angular acceleration (bottom-right) were obtained from the Prevent Biometrics portal and have been processed and transformed to the head's centre of gravity by the proprietary algorithms and the red, green and blue lines are aligned with the X-, Y- and Z-axes of the head's coordinate system.

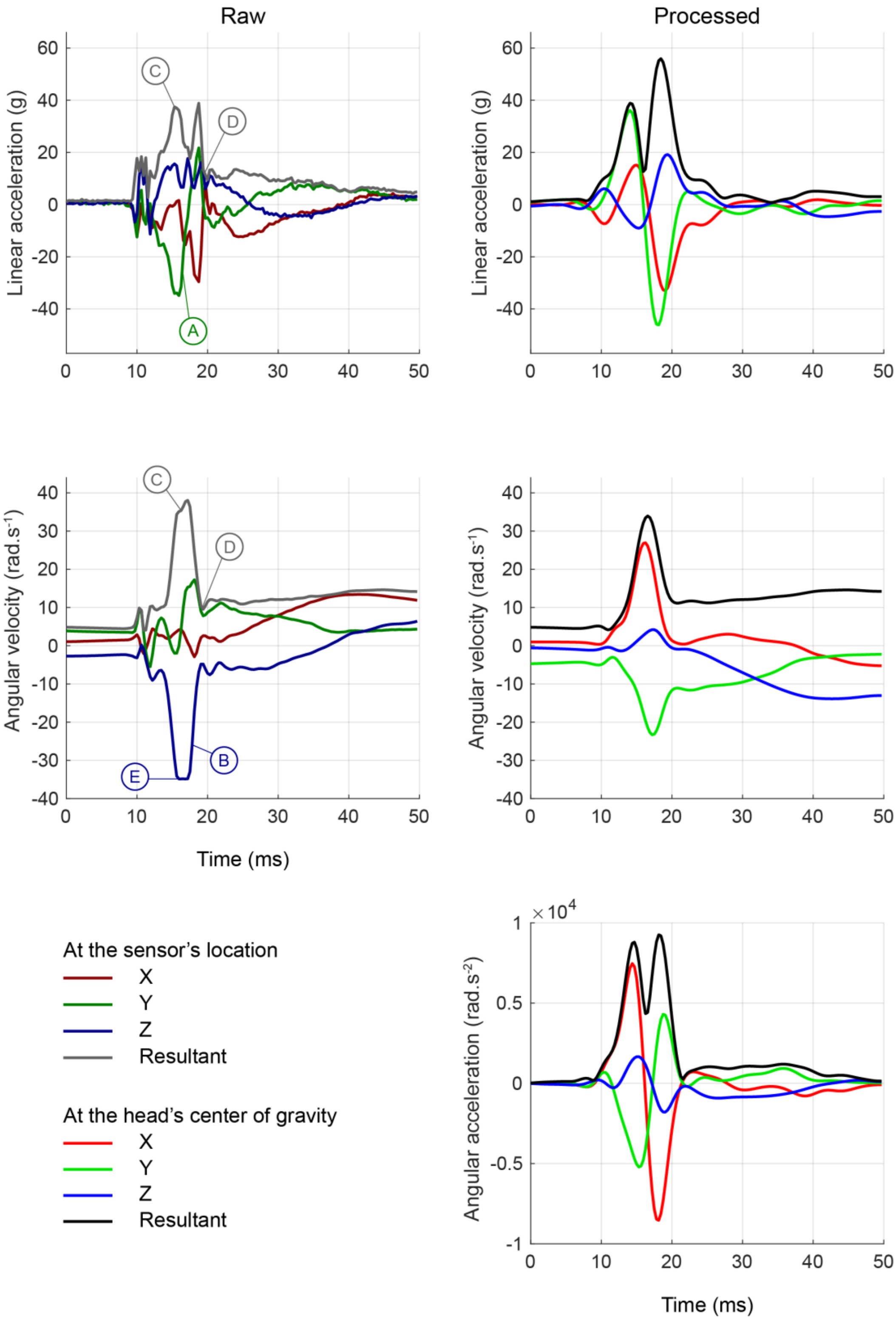


Figure 5.3. Example of a 'bad' signal from the mouthguard. The raw linear acceleration (top-left) and raw angular velocity (middle-left) were provided by Prevent Biometrics and are plotted at the sensor's location. Note the orientation of the sensor's axes is unknown for the raw data. The processed linear acceleration (top-right), angular velocity (middle-right) and angular acceleration (bottom-right) were obtained from the Prevent Biometrics portal and have been processed and transformed to the head's centre of gravity by the proprietary algorithms and the red, green and blue lines are aligned with the X-, Y- and Z-axes of the head's coordinate system. The raw signals display several characteristics of a bad signal: (A) rapid signal inversion on the Y raw linear acceleration, (B) sharp spike on the Z raw angular velocity, (C) peaks in linear acceleration and angular velocity aligned in time, (D) noisy signal suddenly becoming smooth, and (E) saturation of the sensor.

Variable extraction

From the video analysis (see Chapter 4), we extracted the following information:

- the type of impact (definite head impact vs prolonged contact, direct vs. indirect)
- impact location (sensor vicinity, in the face area, location by bins)
- whether the impact was a scoring punch, not a scoring punch, or had not been assessed by the boxing judge
- whether it was associated with a heavy blow reported by the participants

Additionally, the distance between the impact location determined on video and each sensor was calculated from a two-dimension image (Figure 4.5). The cartesian position of each sensor was estimated on the right profile for the skin patch and on both the right and left profiles for the headgear patch and mouthguard. The cartesian location of the impact was approximated on the same image and the distance, in pixels, was calculated. The shortest of the two possible distances (right of left side) for the headgear patch and the mouthguard was selected. Finally, PLA and PAA were calculated from the processed data as the maximum value of the resultant LA or AA, respectively.

Analyses

Raw signals quality

After the raw signals' quality was assessed, the overall proportions of 'good', 'questionable' and 'bad' recordings were calculated, as well as the prevalence of the main types of issues that were observed in the 'bad' recordings (Table 5.1). The proportions and 95% confidence interval (CI) of 'good' recordings were calculated for participants, type of impact and impact location, and represented with forest plots. Differences between participants, types of impact or impact locations were determined significant if the 95% CIs did not overlap.

Association between signal quality and peak kinematics

The distributions of PLA and PAA for the 'good', 'questionable' and 'bad' impacts for all sensors were not normal (confirmed by a Shapiro-Wilk test of normality) and had unequal variances (confirmed by a Levene test). Therefore, we used non-parametric tests to determine how the quality of the signals and the peak kinematics were related. First, the differences in PLA and PAA between the three quality classes were assessed with a Kruskal-Wallis analysis of variance, completed with post-hoc tests using a Dunn-Sidák correction to assess the differences between the individual quality classes. Descriptive statistics were also calculated for all video-verified recordings, all included recordings, and for each quality class.

Next, the differences in PLA and PAA by type of impact (direct vs. indirect, scoring vs. non-scoring punches) and impact location (in the face vs. away from the face, in the sensor vicinity vs. elsewhere) were assessed using Wilcoxon rank-sum tests. The differences between participants or bins of impact location (front, left, right, and back) were evaluated with Kruskal-Wallis analysis of variance and post-hoc tests with a Dunn-Sidák correction. Additionally, the correlation between the distance separating the impact to the sensor and peak accelerations was assessed with a Spearman's rank correlation test. All these analyses were conducted three times, once for each of the following data subsets: all video-verified recordings, all included recordings, and 'good' recordings only. By comparing the results across the data subsets, we were able to determine how the inclusion of 'bad' and 'questionable' recordings could affect the differences in PLA and PAA between types of impact, impact locations and participants.

Comparison of peak accelerations for simultaneous recordings

To compare the kinematics measured by the mouthguard (used as the reference), the skin patch, and the headgear patch, we first considered all the 'good' recordings available for each sensor and assessed the difference in the distribution of PLA and PAA with a Kruskal-Wallis analysis of variance, completed with post-hoc tests with a Dunn-Sidák correction. Next, we assessed the associations between paired 'good' recordings across sensors using Spearman's rank correlation test (e.g., only the recordings that were 'good' for both the mouthguard and the skin patch). Finally, paired comparisons between the sensors' PLA and PAA were assessed using Wilcoxon's rank-sum test.

All analyses were performed using Matlab's statistical functions, and the significance level was set at 0.05.

5.5. Results

Participants' information

Participants' characteristics, location vectors and Euler angles for the alignment of the patches to the head's coordinate system are presented in Table 5.2.

Table 5.2. Participants' anthropometrics and estimated location and orientation of the skin and headgear patches. The head's axes are oriented as follows: X towards the front, Y towards the right, Z towards the bottom.

Participant ID	Sex	Current weight (kg)	Total body height (cm)	Head length (cm)	Head breadth (cm)	Headgear length (cm)	Headgear breadth (cm)
03	F	65.7	163.7	18.0	17.0	19.5	18.5
04	F	61.0	170.4	19.5	16.0	24.0	18.0
05	F	52.0	164.5	17.0	17.0	21.5	19.5
07	F	56.0	176.3	19.0	15.5	24.0	20.5
08	M	88.0	191.5	20.5	15.5	25.0	19.0
09	M	116.0	192.5	20.0	17.0	26.0	21.0
10	M	73.0	169.5	18.0	16.0	N/A	N/A
Mean ± SD		73.1 ± 22.4	175.5 ± 12	18.9 ± 1.2	16.3 ± 0.7	23.3 ± 2.4	19.4 ± 1.2

Table 5.2. (continued)

	Skin patch								Headgear patch							
Participant ID	N	Distance X (cm)	Distance Y (cm)	Distance Z (cm)	N	ψ (°)	ϕ (°)	θ (°)	N	Distance X (cm)	Distance Y (cm)	Distance Z (cm)	N	ψ (°)	ϕ (°)	θ (°)
03	2	4.1 ± 0.0	-8.1 ± 0.4	-2.3 ± 0.3	1-2	7.0	4.0 ± 5.7	10.0	1	10.4	1.2	1.2	1	0	-10	5.0
04	4	4.1 ± 0.4	-8.3 ± 0.7	-2.2 ± 0.6	4	13.0 ± 2.4	-15.1 ± 12.6	5.5 ± 4.2	3	13.3 ± 0.7	-0.4 ± 0.7	0.5 ± 0.7	3	3.3 ± 5.8	-8.0 ± 5.3	0 ± 0
05	4	3.7 ± 0.1	-8.1 ± 0.9	-2.1 ± 0.7	4	8.0 ± 2.4	-4.4 ± 5.9	1.3 ± 2.5	3	11.9 ± 0.2	0.9 ± 0.2	-1.8 ± 1.6	3	0 ± 0	-14.0 ± 5.6	0 ± 0
07	2-3	3.6 ± 0.3	-7.6 ± 0.4	-1.7 ± 0.3	1-3	10.0	-22.5 ± 11.1	0 ± 0	2	14.3 ± 0.2	0.2 ± 1.2	0.1 ± 0.8	2	-5.5	-5.0 ± 0	0 ± 0
08	1	4.1	-7.4	-2	1	0	-11.0	5.0	1	11.9	0.7	1.0	1	0	-3.0	0
09	3	4.2 ± 0.1	-8.3 ± 0.2	-1.1 ± 0.3	3	1.3 ± 2.3	-14.8 ± 7.6	0 ± 0	2-3	13.6 ± 2.1	0.2 ± 1.1	-2.4 ± 0.1	2-3	-3.3 ± 2.9	-7.5 ± 3.5	-2.5 ± 3.5
10	1-2	3.7 ± 0.2	-7.6	-2.0 ± 0.8	1-2	0	-20.0 ± 1.4	0	0	*	*	*	1-2	3.5 ± 4.9	-10.0	0
Mean ± SD		3.9 ± 0.2	-7.9 ± 0.4	-1.9 ± 0.4		5.6 ± 5.2	-12.0 ± 9.2	3.1 ± 3.9		12.5 ± 1.4	0.5 ± 0.5	-0.2 ± 1.5		-0.3 ± 3.3	-8.2 ± 3.6	0.4 ± 2.2

Participants were identified by the number of their mouthguard (MG03 to MG10, with MG01 and MG02 kept as back-up and MG06 having withdrawn before the start of the study). The numbers in columns N represent the number of photographs that were used to extract distances and angles; the numbers vary from one measurement to another because of what could be discerned on the images. ψ, ϕ, θ are the Euler angles about the x'', y'-, and z-axes, respectively, estimated from the photographs and decomposed in the order z-y'-x''.
*The headgear dimensions were unavailable for participant #10, therefore the average distances and angles for all other participants were used for the transformation of the linear acceleration to the head's centre of gravity.

Raw signals quality

A total of 442 matched events for each sensor were visually assessed for quality. The overall proportion of ‘good’ sensor recordings was 53, 20, and 26% for the mouthguard, skin patch, and headgear patch, respectively (Figure 5.4). Mouthguard recordings assessed as ‘bad’ largely presented a rapid signal inversion and/or a sharp spike on the LA traces (85% of ‘bad’ recordings) compared with AV traces (54%). The patches’ issues were mostly driven by the AV signals, which recorded a rapid signal inversion and/or a sharp spike for 90% and 95% of the ‘bad’ recordings for the skin and headgear patches, respectively (vs. 35% and 26% for the LA signals). Matching peaks in LA and AA were observed for 74, 60, and 68% of the ‘bad’ mouthguard, skin patch, and headgear patch recordings, respectively. The mouthguard’s accelerometers reached their saturation point (±200 g) on at least one channel for 5% of the ‘bad’ recordings, while the angular rate sensors (±35 $rad.s^{-1}$) did so in 16% of the ‘bad’ recordings. Saturation of the CSx angular rate sensors (±70 $rad.s^{-1}$) was also reached for 5% and 1% of the ‘bad’ recordings for the skin and headgear patch, respectively (less than 2% for the accelerometer).

The proportion of ‘good’ recordings statistically differed from the overall proportion for one or two participants for the patches (Figure 5.5). The type of impact did not significantly affect the proportions of ‘good’ recordings on either sensor. The mouthguard and the skin patch had significantly lower proportions of ‘good’ recordings for the impacts that landed in their vicinity (30% vs 61% for the mouthguard, 4% vs. 22% for the skin patch). The number of impacts in the headgear patch vicinity was only 5, all of them classified as ‘bad’. Being hit in the face (vs. elsewhere on the head) led to proportions of ‘good’ recordings being lower for the mouthguard (22% when hit in the face vs. 60% when hit away from the face), similar for the skin patch (21 vs. 20%), and higher for the headgear patch (40 vs. 22%). Lower proportions of ‘good’ recordings were also observed for impacts occurring to the front (vs. the sides) for the mouthguard, to the right side (vs. the front and left side) for the skin patch, and to the sides and back (vs. the front) for the headgear patch.

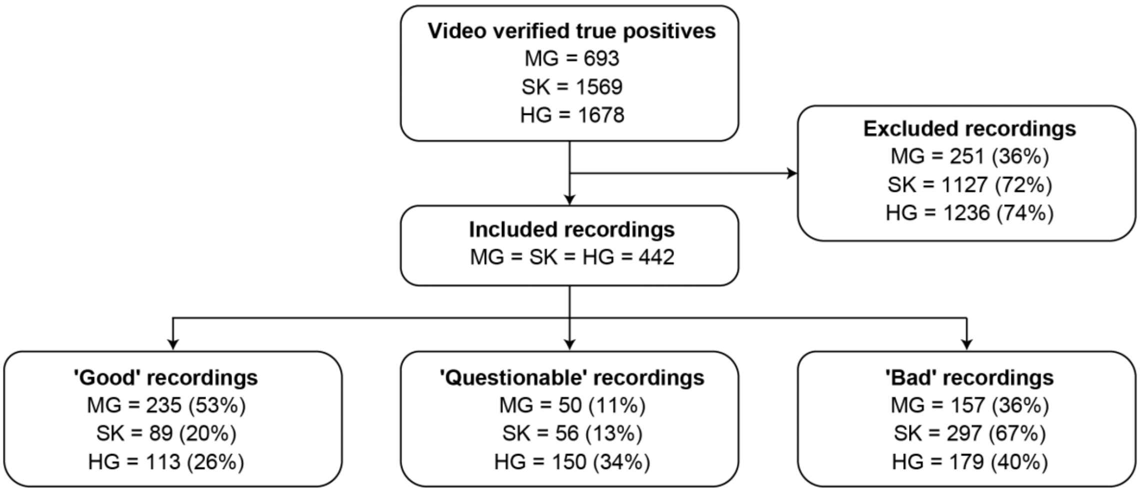


Figure 5.4. Number of recordings included for the data quality assessment and result of the assessment. Included recordings comprise only recordings that could be matched across all three sensors and to a specific definite head impact or prolonged contact observable on video. MG: mouthguard, SK: skin patch, HG: headgear patch.

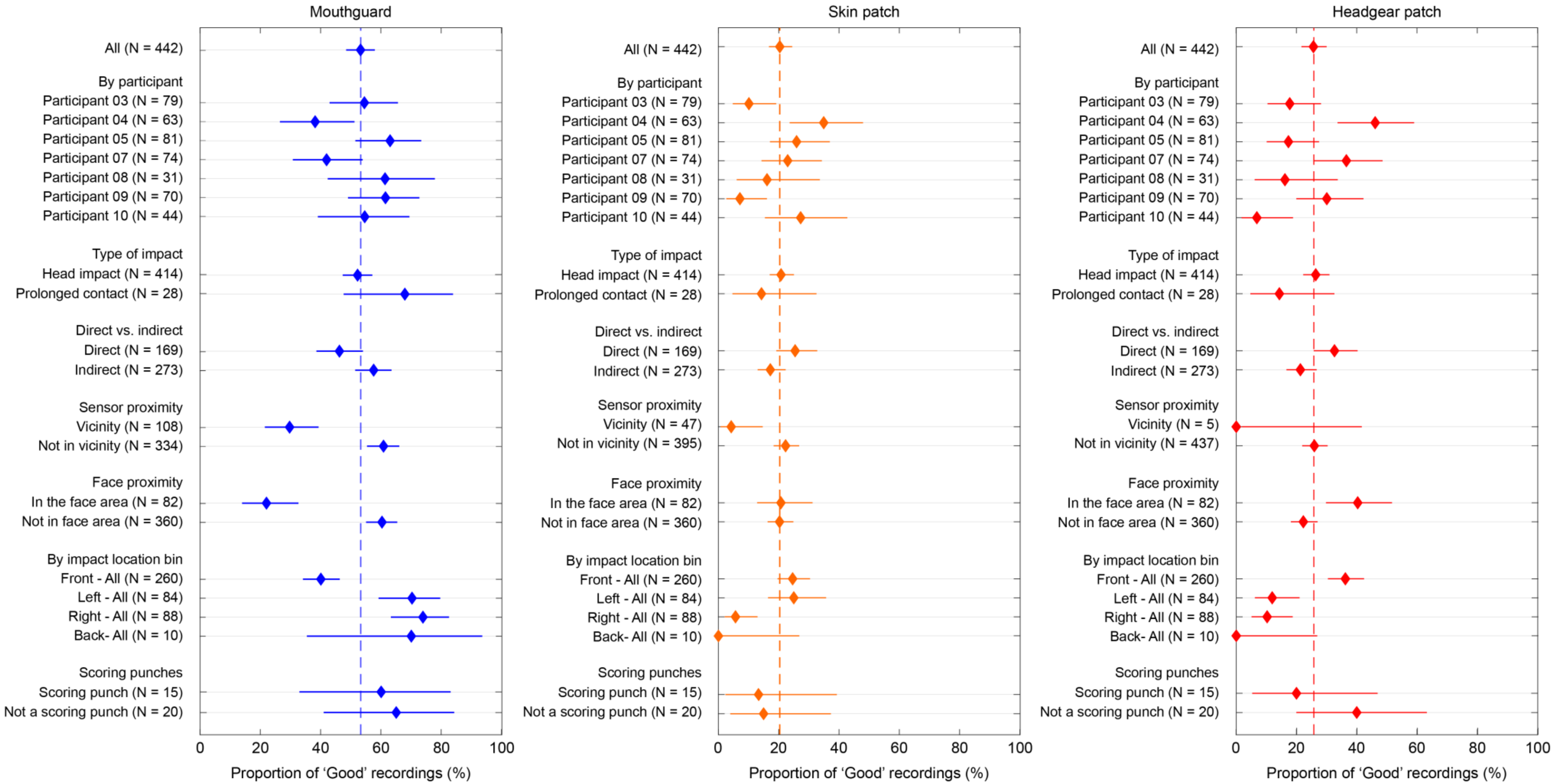


Figure 5.5. Proportion of 'good' recordings for the mouthguard (left), skin patch (middle) and headgear patch (right), by participant, type of impact and impact location (as determined from the video analysis). The diamonds represent the proportion of ‘good’ recordings for the category and the lateral bars the 95% confidence intervals. The coloured dashed vertical lines represent the overall proportion for each sensor. Statistical significance is assumed when the 95% confidence intervals do not overlap.

Association between signal quality and peak kinematics

The ‘good’ recordings presented significantly lower PLA and PAA than the ‘questionable’ recordings ($p < 0.05$, apart from the PLA measured by the skin patch: $p = 0.31$), and than the ‘bad’ recordings for all sensors ($p < 0.001$)(Figure 5.6, Table 5.3). The ‘questionable’ recordings also showed significantly lower PLA and PAA than the ‘bad’ recordings (all sensors, $p < 0.001$).

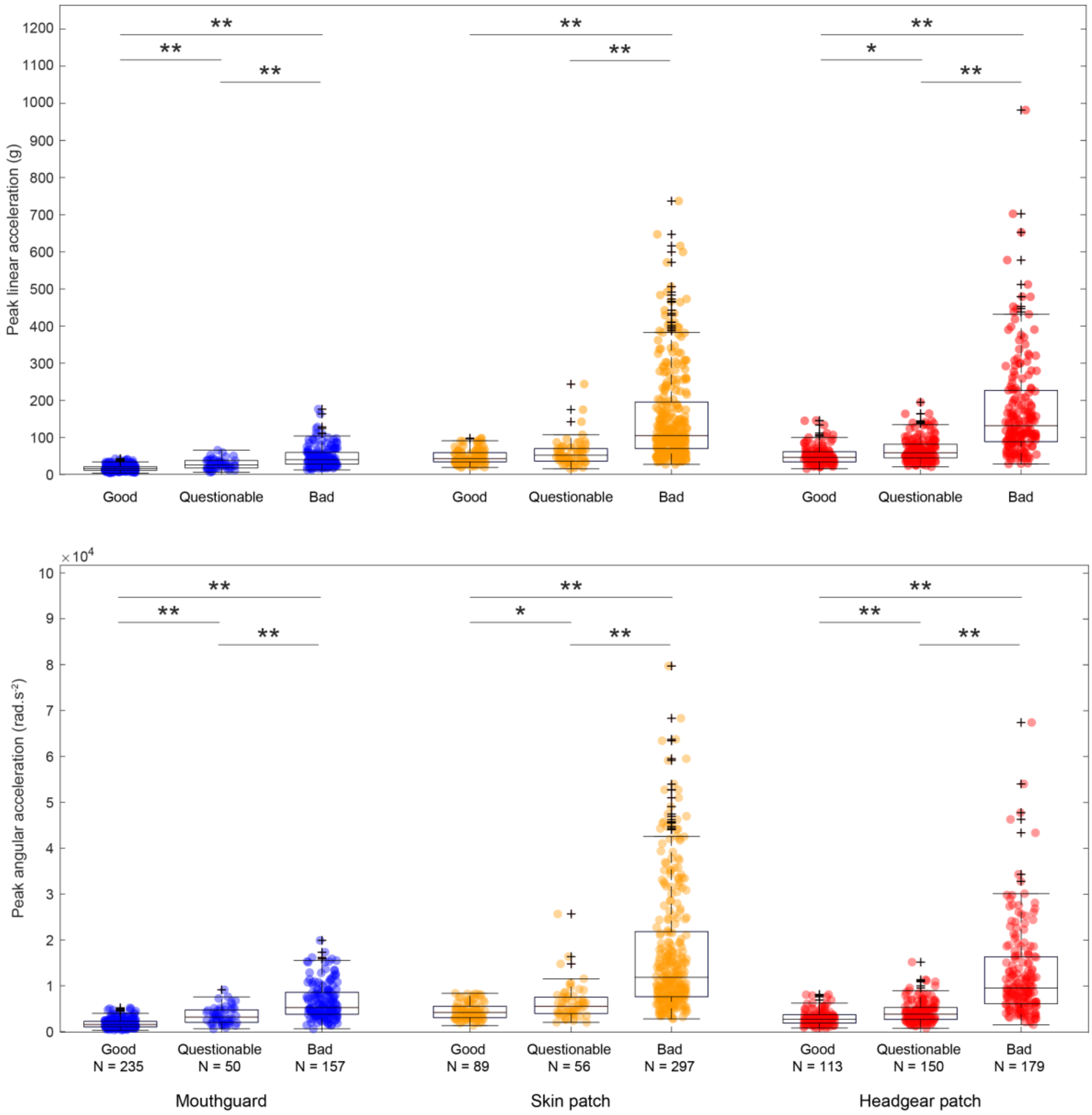


*Figure 5.6. Distribution of the peak linear acceleration (top) and peak angular acceleration (bottom) at the head’s centre of gravity by quality of the recording assessed by visual analysis of the raw data. Each dot (blue for the mouthguard, orange for the skin patch and red for the headgear patch) represent one data point. On each box, the central line indicates the median, and the bottom and top edges of the box indicate the 25th and 75th percentiles, respectively. The whiskers extend to the most extreme data points not considered outliers, and the outliers (any value that is more than 1.5 times the interquartile range away from the bottom or top of the box) are plotted individually using the '+' symbol. The horizontal lines and asterisks above the plots indicate a significant difference between the mean ranks of the two groups, determined from a Kruskal-Wallis analysis of variance, completed with post-hoc tests with a Dunn-Sidák correction (*p < 0.05, **p < 0.001).*

Table 5.3. Descriptive statistics for the resultant peak linear and angular accelerations measured by the mouthguard, skin patch, and headgear patch, at various steps of the inclusion process.

	Mouthguard			Skin patch			Headgear patch		
	N	PLA (g)	PAA (rad.s^{-2})	N	PLA (g)	PAA (rad.s^{-2})	N	PLA (g)	PAA (rad.s^{-2})
Video-verified	693			1569			1678		
Median		22.14	2573.62		61.38	6921.86		54.98	3590.46
IQR		14.58 - 38.41	1508.55 - 5016.81		38.81 - 112.21	4001.15 - 12807.65		37.06 - 96.33	2199.32 - 6647.95
Mean ± SD		30.43 ± 23.4	3901.23 ± 3432.72		96.74 ± 95.25	10676.43 ± 10697.27		86.85 ± 98.31	5889.16 ± 7142.13
Range		4.47 - 176.54	414.44 - 20000.55		11.54 - 736.06	538.77 - 79717.18		11.87 - 980.93	644.99 - 67392.12
Included recordings	442			442			442		
Median		21.85	2556.49		79.68	8231.02		71.88	4674.57
IQR		14.6 - 36.99	1532.62 - 4854.87		48.73 - 139.33	4911.96 - 15806.52		46.77 - 119.12	2890.31 - 8514.47
Mean ± SD		29.73 ± 23.16	3769.21 ± 3320.39		120.25 ± 114.22	13301.8 ± 12912.65		106.64 ± 106.64	7337.27 ± 7859.45
Range		4.47 - 176.54	414.44 - 20000.55		14.87 - 736.06	1331.64 - 79717.18		14.93 - 980.93	752.46 - 67392.12
'Good' recordings	235			89			113		
Median		16.32	1685.35		41.54	4167.76		45.74	2702.53
IQR		11.77 - 21.37	1129.75 - 2358.9		33.34 - 58.12	3040.91 - 5416.03		33.41 - 60.98	1860.7 - 3691.08
Mean ± SD		17.63 ± 7.96	1851.74 ± 952.1		47.41 ± 19.67	4423.76 ± 1811.04		50.95 ± 25.43	3033.15 ± 1630
Range		4.47 - 43.01	414.44 - 5292.28		18.66 - 97.42	1331.64 - 8379.74		14.93 - 144.83	798.72 - 8098.76
'Questionable' recordings	50			56			150		
Median		26.66	3283.86		51.61	5587.83		58.24	3820.64
IQR		18.07 - 38.06	2120.21 - 4854.87		35.75 - 70.62	3969.16 - 7653.22		44.21 - 81.11	2670.11 - 5247.74
Mean ± SD		28.23 ± 13.38	3655.93 ± 1888.83		59.93 ± 38.2	6381.28 ± 3940.58		66.69 ± 31.98	4339.48 ± 2336.32
Range		6.26 - 65.99	726.85 - 9222.96		14.87 - 242.84	2010.23 - 25661.83		20.01 - 193.98	752.46 - 15167.53
'Bad' recordings	157			297			179		
Median		40.25	5342.07		104.09	11838.93		130.89	9527.49
IQR		28.11 - 59.86	3899.06 - 8700.93		69.41 - 194.66	7612.3 - 21813.99		88.09 - 225.98	6079.13 - 16316.92
Mean ± SD		48.31 ± 28.22	6675.37 ± 3826.72		153.45 ± 125.14	17267.1 ± 14000.63		175.28 ± 137.34	12566.5 ± 9998.31
Range		12.79 - 176.54	713.32 - 20000.55		26.89 - 736.06	2789.85 - 79717.18		27.91 - 980.93	1506.7 - 67392.12

PLA: peak linear acceleration, PAA: peak angular acceleration, IQR: interquartile range, SD: standard deviation.

Direct vs Indirect

When all video-verified recordings were considered, direct impacts led to higher PLA and PAA than indirect impacts for the mouthguard, but lower PLA and PAA for the skin and headgear patches (all $p < 0.001$). These differences were maintained when considering matched recordings only ($p < 0.007$), but were only significant for the PAA measured by the mouthguard when the dataset was rarefied to the 'good' recordings (median of 1818.23 rad.s^{-2} [IQR: 1255.50 - 2621.61] for direct impacts vs. 1558.67 rad.s^{-2} [1070.42 - 2155.18] for indirect impacts, $p = 0.007$)(all other $p > 0.12$).

Impact location

For all sensors when 'good', 'questionable' and 'bad' recordings were considered, the median PLA and PAA were significantly higher for impacts landing in the vicinity of the sensor ($z > 2.48$, $p < 0.013$). There were weak but significant negative linear correlations between the distance separating the impact location to the sensor and the peak metrics for all sensors ($p < 0.001$, R^2 = 0.06 - 0.19)(Figure 5.7). The median PLA and PAA also differed when impact location was assessed by bins (Kruskal-Wallis analysis of variance, $p < 0.001$), with differences consistent between PLA and PAA for each sensor, but specific to each sensor (Figure 5.8). However, in each case, every difference and correlation lost their significance when the 'bad' and 'questionable' recordings were removed.

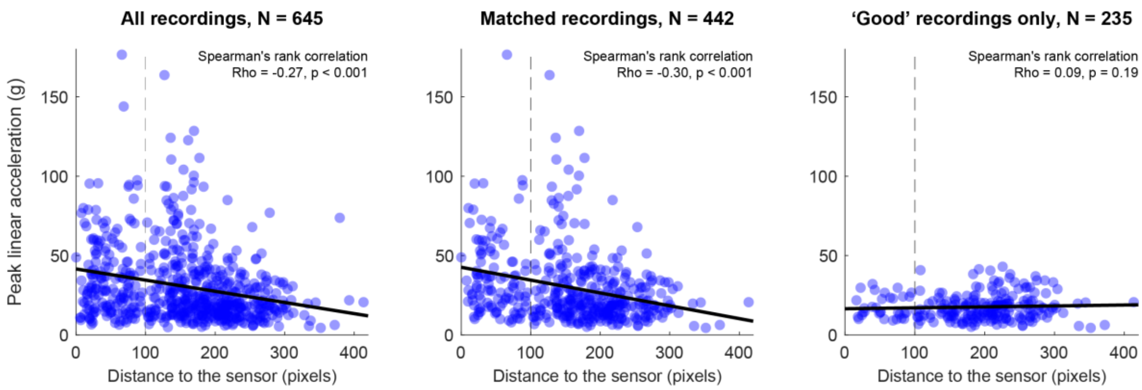


Figure 5.7. Peak linear acceleration as a function of the distance between the location of the impact on the head and the estimated location of the mouthguard, for all recordings (left), all matched recordings (centre) and 'good' recordings only (right). The thick black line represents the regression line of a linear model. The vertical dashed line at 100 pixels represents the boundary of the dichotomous variable of sensor vicinity: all data points of the left of this line are estimated to be in the vicinity of the mouthguard.

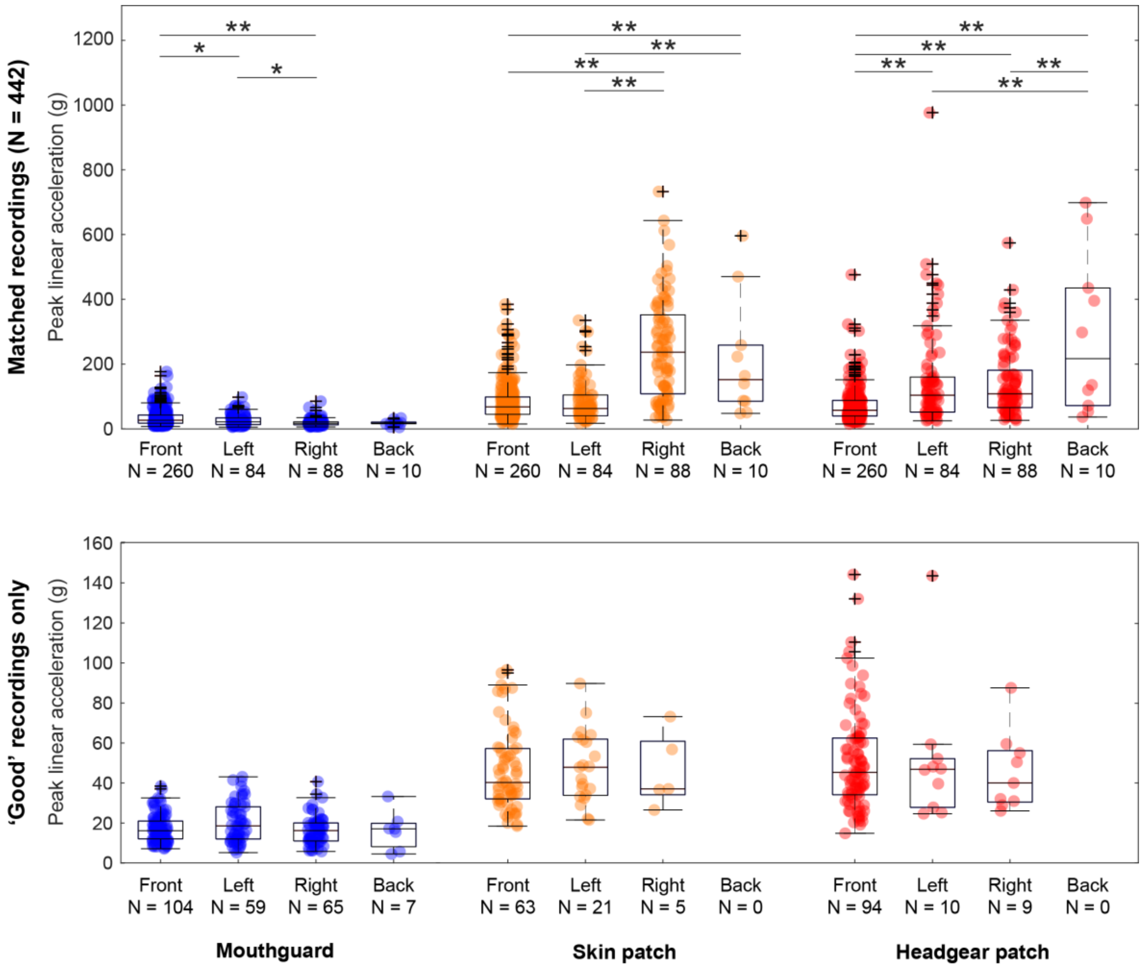


*Figure 5.8. Distribution of the peak linear acceleration measured by the mouthguard (left), skin patch (middle), and headgear patch (right) across the four impact location bins, as determined by the video analysis. The top panel comprises all matched recordings, the bottom panel only the recordings assessed as 'good' for each sensor (note the different scale). On each box, the central line indicates the median, and the bottom and top edges of the box indicate the 25th and 75th percentiles, respectively. The whiskers extend to the most extreme data points not considered outliers, and the outliers (any value that is more than 1.5 times the interquartile range away from the bottom or top of the box) are plotted individually using the '+' symbol. The horizontal lines and asterisks above the plots indicate a significant difference between the mean ranks of the groups, determined from a Kruskal-Wallis analysis of variance, completed with post-hoc tests with a Dunn-Sidák correction (*p < 0.05, **p < 0.001).*

Scoring punches and heavy blows

The review of sparring videos by the certified judge in the previous chapter had yielded 17, 42, and 38 mouthguard, skin patch, and headgear patch recordings, respectively, that were associated with a scoring punch, along with 33, 75, and 82 recordings that were associated with a head impact that was not a scoring punch. After matching recordings and removing the 'questionable' and 'bad' recordings, the final numbers were 9, 2, 3 for the scoring punches and 13, 3, 8 for the non-scoring punches for the mouthguard, skin patch and headgear patch, respectively (Figure 5.9). There was no significant difference in PLA and PAA between scoring and non-scoring punches for any sensor, at any level of inclusion (all, matched or 'good' recordings only: $p > 0.49$ for PLA and > 0.06 for PAA), although the difference was close to being significant for PAA measured by the mouthguard with ‘good’ recordings only (scoring punches (N = 9): 2124.67 rad.s^{-2}, non-scoring punches (N = 13): 1830.36 rad.s^{-2}, p = 0.062).

The two heavy blows that could be identified on video were from rear hooks to the left ear area (i.e., away from all sensors), and recordings from all three sensors were matched. Only one mouthguard recording was determined ‘good’ and showed a PLA of 41.4 g, a PAV of 21.7 rad.s^{-1} and a PAA of 5292.3 rad.s^{-2}. This particular impact led to the highest PLA and PAA recorded on a ‘good’ mouthguard recording for this participant, the closest peaks being at 32.60 g and 2760 rad.s^{-2} (Figure 5.9, MG04).

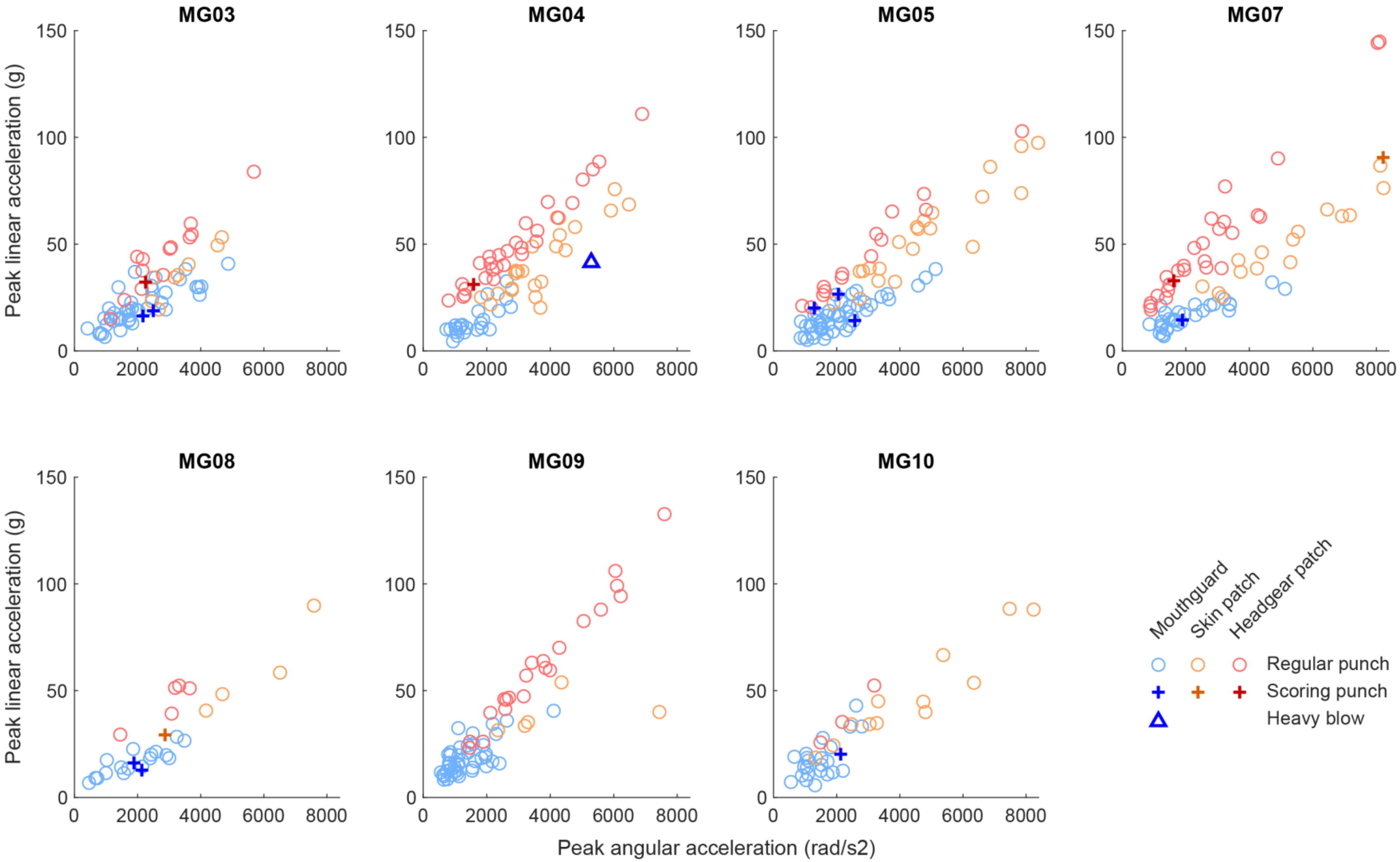


Figure 5.9. Peak linear and angular acceleration of the 'good' recordings by type of punch for each participant. Note: only 30% of the overall dataset has been assessed with regards to scoring punches, so the number of scoring punches is likely underestimated.

Comparison of peak accelerations for simultaneous recordings

Considering all the ‘good’ recordings available for each sensor, the median PLA was lower for the mouthguard than for both patches ($p < 0.001$) and the median PAA differed between all three sensors ($p < 0.001$)(Figure 5.10).

We were able to compare 38 ‘good’ mouthguard and skin patch recordings, 50 ‘good’ mouthguard and headgear patch recordings, and 32 ‘good’ skin and headgear patches recordings (Figure 5.11). The recordings were assessed as ‘good’ for all three sensors in only 14 instances (example in Figure 5.12). There was a positive correlation between the headgear patch and the mouthguard for the peak angular acceleration (Spearman’s $p = 0.043$, rho = 0.288). There was no correlation between any sensors for the peak linear acceleration, nor for the angular acceleration between the mouthguard and the skin patch and between the skin patch and the headgear patch (Spearman’s $p > 0.36$). In addition to the absence of correlation, the medians of the peak values reported by the patches were two to three times larger than the values reported by the mouthguard ($p < 0.001$)(Table 5.4). The median PLA was similar between the two patches ($p = 0.52$), but the median PAA was significantly higher for the skin patch than the headgear patch ($p < 0.001$).

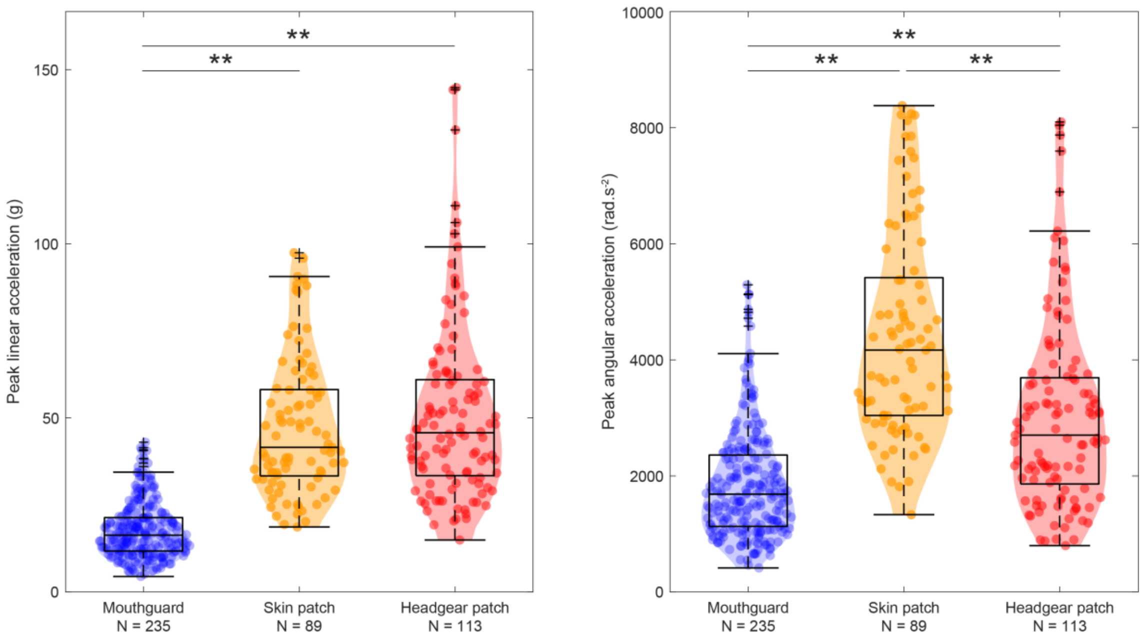


Figure 5.10. Distribution of peak linear and angular accelerations for the 'Good' recordings. Each dot (blue for the mouthguard, orange for the skin patch and red for the headgear patch) represent one data point. On each box, the central line indicates the median, and the bottom and top edges of the box indicate the 25th and 75th percentiles, respectively. The whiskers extend to the most extreme data points not considered outliers, and the outliers (any value that is more than 1.5 times the interquartile range away from the bottom or top of the box) are plotted individually using the '+' symbol. The shaded areas represent the distribution of the data and were smoothed with a kernel density function using the Matlab function ksdensity *with default settings. The horizontal lines and asterisks above the plots indicate a significant difference between the mean ranks of the two groups, determined from a Kruskal-Wallis analysis of variance, completed with post-hoc tests with a Dunn-Sidák correction (**$p < 0.001$).*

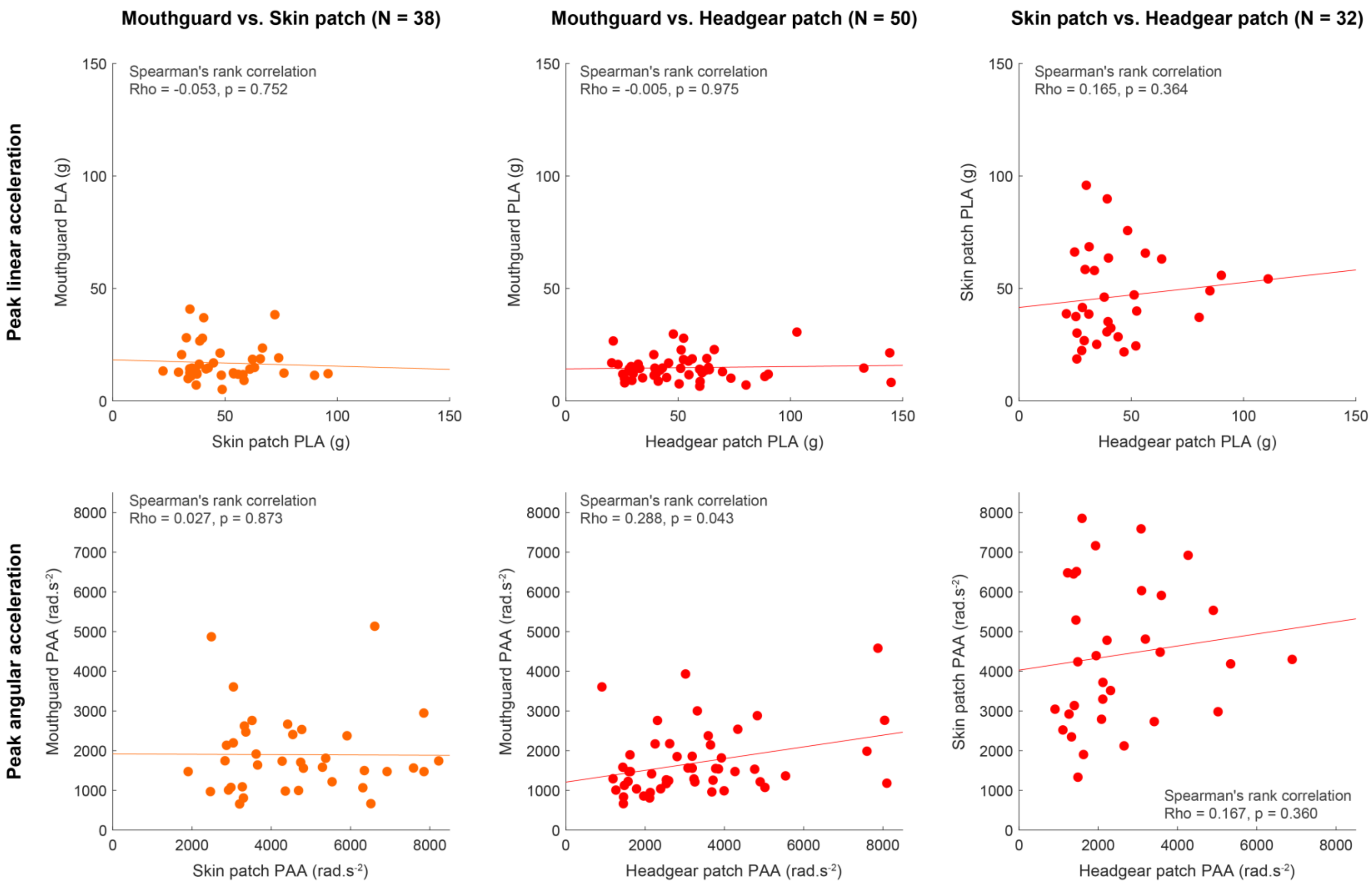


Figure 5.11. Correlations between the peak kinematics measured by the three sensors (top: peak linear acceleration (PLA), bottom: peak angular acceleration (PAA)). Each dot represents a recording that could be matched between the two sensors and to a specific event identifiable on video. The results from Spearman's rank correlations are reported on each subplot. The coloured lines represent the linear regression line.

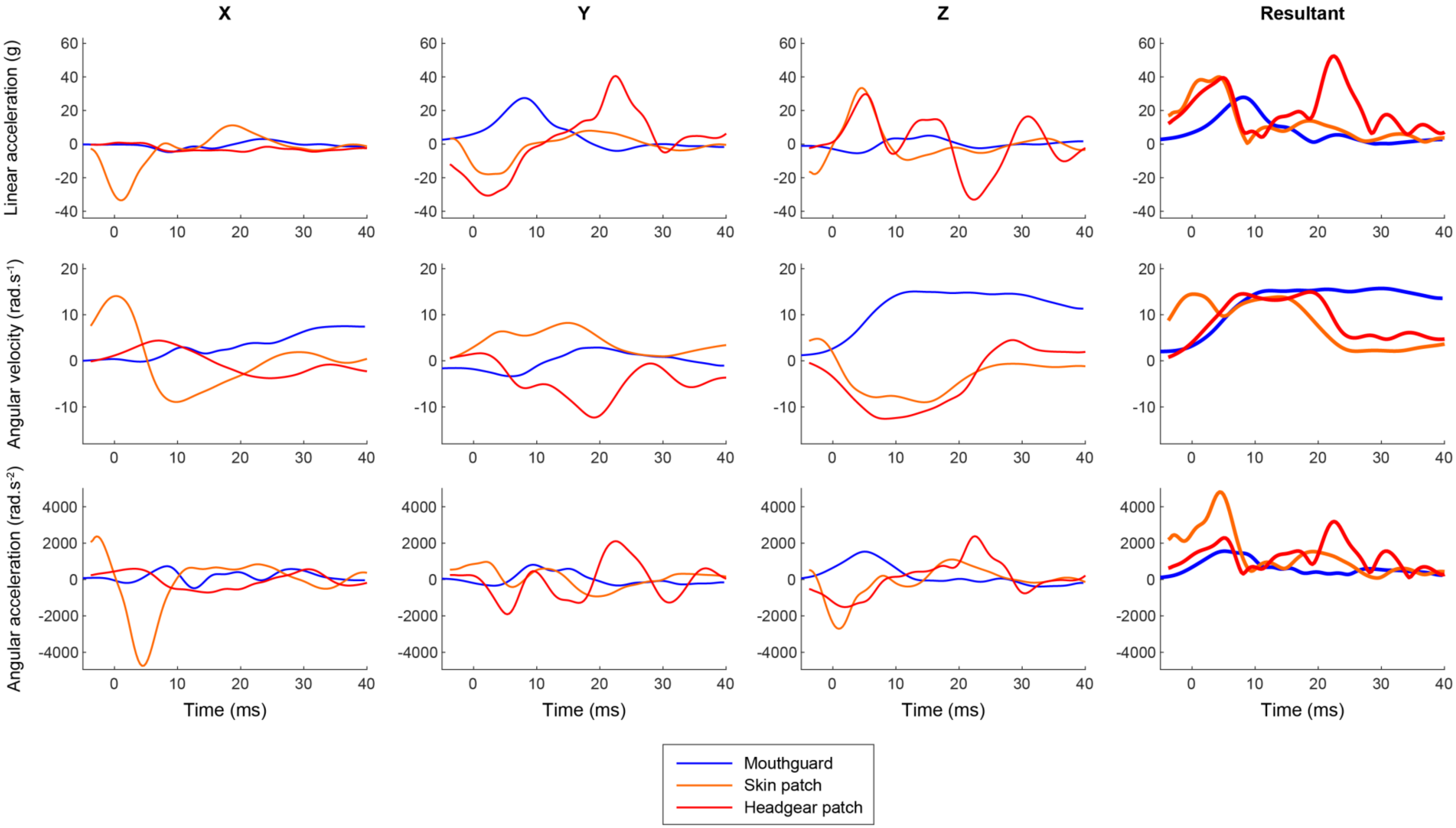


Figure 5.12. Comparison of the individual components and resultant of exemplar matched (synchronous) recordings from the mouthguard (blue), skin patch (orange) and headgear patch (red). All three recordings have been assessed as being 'good' based on the visual analysis of the raw data, have been resolved to the same reference frame, and have been aligned on the crossing of the 10-g threshold by any raw linear acceleration signal.

Table 5.4. Medians and interquartile ranges for peak linear and angular acceleration for the groups of 'good' recordings being compared between two sensors, and results of Wilcoxon rank-sum tests for each paired comparison.

Mouthguard	Skin patch	Headgear patch	N	z	p
Peak linear acceleration (g)					
14.18 [11.93 - 19.14]	43.66 [35.25 - 61.05]	-	38	7.04	< 0.001
14.00 [10.86 - 16.93]	-	50.78 [32.87 - 62.77]	50	8.34	< 0.001
-	40.75 [30.43 - 60.78]	39.21 [29.24 - 51.66]	32	-0.64	0.52
Peak angular acceleration (rad.s^{-2})					
1673.05 [1089.40 - 2407.12]	4384.23 [3200.80 - 5911.45]	-	38	6.56	< 0.001
1473.01 [1169.95 - 1986.28]	-	3050.12 [1963.05 - 3925.56]	50	5.53	< 0.001
-	4269.46 [2951.32 - 5971.88]	2098.49 [1444.27 - 3301.57]	32	-4.10	< 0.001

5.6. Discussion

This observational study had three aims: (1) to assess the quality of the raw kinematic traces measured by a mouthguard, a skin patch, and a headgear patch, by determining the types of issues present and their prevalence; (2) to determine the association between signal quality and peak kinematics; (3) to compare the peak accelerations measured by the three sensors using only recordings of acceptable quality. To achieve these objectives, we focused our analyses on acceleration events that were matched for all three sensors and associated with a specific video-verified head impact. Upon examination of the raw signals, between 20 and 53% of recordings for each sensor satisfied our quality criteria. Most recordings that did not meet our criteria contained rapid signal inversions and/or sharp spikes in the linear acceleration and/or the angular velocity signals. For all sensors, the prevalence of 'bad' recordings was substantially higher for impacts landing in the proximity of the sensor. Our findings showed that the low quality of signals was associated with significantly larger peak kinematics, which in turn introduced significant differences between impact types and locations. Ultimately, when the problematic recordings were excluded from the analyses, we found that the patches generally measured higher peak linear and angular accelerations than the mouthguard and that there was little to no association between the sensors' measurements for simultaneous recordings.

Raw signals quality

The quality assessment of the raw signals revealed that large proportions of acceleration events presented anomalies that appear to be unrelated to true skull motion (74-80% for the patches, 47% for the mouthguard). Similar issues have been reported with other mouthguards[389, 418] and skin patches.[343]

The rapid signal inversions and sharp spikes observed in 85% of the 'bad' mouthguard LA traces indicate that the mouthguard either suddenly decoupled from the teeth, got snapped into place, or moved from one decoupled position to another. The change in coupling state is consistent with a participant suddenly clenching their teeth upon receiving a punch or being hit directly on the jaw, which increased the risk of anomalies (from 40 to 70% risk of 'bad' or 'questionable' recordings). Having deployed mouthguards in Mixed Martial Arts, Tiernan et al. excluded all such events where the impact landed directly on the mouthguard, as they observed sharp spikes in the signals.[389] A laboratory investigation revealed that a slack jaw, in opposition to clenched, can introduce errors in the kinematic signals and reduces the validity of the mouthguard's measurements.[222] Jaw slackness is not something that can be controlled *in-vivo* and might have contributed to differences in the proportion of good-quality signals between participants. Factors that may reduce the influence of a slack jaw are mouthguard design (e.g., better isolation of the sensors)[222] and quality of fit. Indeed, laboratory testing found that the accuracy of custom-fit mouthguards was more consistent than that of boil-and-bite mouthguards.[232] More research is needed to objectively assess mouthguard fit and how it affects the overall signal quality.

The low overall proportion of good-quality signals (20%) found in the skin patch dataset is consistent with Rooks et al.'s report of 11% of 'good' skin patch recordings from their use in military combat sparring.[343] In our study, the poor quality of the recordings was particularly visible in the angular data. This finding aligns with multiple reports of the poorer validity of skin patches for angular measurements than linear measurements,[198, 367, 388] resulting from poor coupling with the skull. We also observed partial and full dislodgement of the skin patch during data collection because of intense sweating. The skull/patch coupling was therefore presumably worse in our study than in cadaver or soccer headings studies, which may also explain the sensitivity to impact location (the quality of the recordings was particularly low for the impacts landing in the vicinity of the sensor). We also hypothesise that many punches caused the headgear to rub directly on the patch. It is possible that a stronger adhesive, the use of strapping tape on the patch and around the head, or no headgear could have helped keeping the skin patch in place and reducing the proportion of 'bad' recordings.

The proportion of 'good' recordings was also low for the headgear patch. We observed that impacts in the proximity of the patch, and more generally to the headgear (i.e., not directly on the face), increased the risk of anomalies in the kinematic traces. This finding accords with Wu et al.'s observations of sharp spikes in skull cap sensor signals when the cap was directly impacted by the soccer ball during headings.[418] In our study, the headgear patch was well coupled to the headgear and the motion it measured was primarily that of the headgear, rather than that of a loose sensor on the headgear. Furthermore, the duration of the spikes in the 'bad' data appeared generally longer than those observed for the mouthguard and skin patch, consistent with motion of an object with more inertia than the patch itself, but less inertia than a head. Overall, our results suggest that the headgear patch measures the independent motion of the headgear rather than the motion of the skull. A tighter fit of the headgear on the head may have resulted in higher proportions of good-quality recordings.

Association between signal quality and peak kinematics

Peak linear and angular accelerations for the 'questionable' and 'bad' recordings were higher than for the 'good' recordings. Any metric or analysis derived from the kinematic traces (e.g., Head Injury Criterion, impact location, injury risk curves) would also be affected by the presence of 'bad' recordings. We also observed significant differences in PLA and PAA between impact types and impact locations when recordings were included independent of quality. Such differences were founded on the interaction between larger proportions of 'bad' recordings and increased magnitude of such recordings and may have led to erroneous conclusions, had we not excluded recordings based on quality. Consequently, there is a risk that differences between impact locations or impact mechanisms reported in other studies were confounded by recordings of poor quality.

Comparison of peak accelerations for simultaneous recordings

From the correlation and comparison analyses between the sensors, we conclude that the skin and headgear patches are not accurate because their measurement differs from the mouthguard's reference value. The patches may not be valid either, i.e., they do not measure skull motion alone.

Our results echo those of other studies that showed either little or no association between mouthguard, xPatch and skull-cap sensor measurements during soccer headings,[418] or weak correlations between HIT System and xPatch measurements during American football games (R^2 = 0.19 for PLA; 0.02 for PAA).[227] Wu et al. also reported 15- and 50-g over-estimations of PLA by the skin patch and skull-cap sensors, respectively (261 and 638% of the mouthguard measurements, compared with 308 and 363% in the present study for the skin patch and headgear patch), as well as 2500- and 4300-rad.s^{-2} over-estimations of the PAA (933 and 1533%, compared with 262 and 207%).[418] Lennon reported that the xPatch and the HIT System measured similar average PLA but the skin-based device reported significantly higher PAA than the helmet-based device (339% higher compared with 203% higher in the present study for the skin patch compared with the headgear patch).

While the CSx sensors proved to be as valid and accurate as other devices in the laboratory (see Chapter 3), they demonstrated important limitations in this *in-vivo* study. The weak associations and large over-estimations of the measurements by the skin- and headgear-based devices relative to the mouthguard are consistent with the decreased validity observed when the sensors are assessed in biofidelic conditions. Indeed, the presence of soft tissue in live experiments or post-mortem tests, as well as a more realistic helmet fit, have led to levels of sensor validity well below what was observed from standard anthropomorphic device testing.[180, 184, 219, 367, 418] These findings led to the hypothesis that skin- and headgear-based devices (helmets, patches, headbands, skull caps) are not rigidly coupled with the skull when used *in-vivo*; our results support this hypothesis. In addition, the sensors' validity may be affected by the location, direction, velocity, and magnitude of the impact (see Chapter 3), and the *in-vivo* environment results in greater variability than what can be tested in the laboratory. Nonetheless, several limitations related to the sensors and the processing of the kinematics signals may have influenced our findings (see Limitations below).

Implications

We reason that the loose coupling between the sensors and the skull resulted in large numbers of recordings showing anomalies in the linear acceleration and angular velocity signals. Such recordings are not representative of skull motion alone and we recommend not using the kinematic data of these recordings in the analysis of exposure and injury risks.

While Tiernan et al. excluded all impacts occurring in the vicinity of the mouthguard,[389] our results showed that this would have excluded 32 valid events (30% of the recordings landing in the vicinity of the mouthguard), but still have left 97 'bad' recordings to be included (29% of the recordings landing away from the mouthguard). Rooks et al. have demonstrated that better algorithms can be developed to improve the identification of 'bad' recordings,[343] and improvements can be expected from both manufacturers and research teams in the future. In the meantime, we recommend adding a quality assessment step to the data cleaning process to ensure that only the kinematics representing head motion are included in the analyses of exposure. However, the visual quality assessment requires the raw data, which is not currently available from many commercial systems.

All three sensors tested in this study showed large proportions of acceleration events that suggest they were not always well coupled to the skull. All three sensors were also highly sensitive to the location of the impact, with impacts landing in the proximity of the sensor worsening the decoupling. The patches were substantially worse than the mouthguard with proportions of questionable and bad recordings reaching 74-80%. Even for recordings satisfying our quality criteria, it appears that the patches, in their current configuration, recorded a motion that was seemingly unrelated to the motion recorded by the mouthguard. While the sensors may perform well in laboratory testing with anthropomorphic devices, our results suggest that the kinematics measured by both patches in our study were unreliable and invalid (i.e., they do not measure head motion). We would therefore advise extreme caution if using skin or headgear patches to appraise the magnitude of exposure in boxing sparring.

The kinematics measured by the mouthguard appeared more reliable (the risk of decoupling was lower than for the patches) and were assumed to be valid for recordings meeting our quality criteria. However, half of the recordings cannot be included in analyses of impact magnitudes due to decoupling-related quality issues. Therefore, in the ongoing investigation of the effects of repetitive head impacts, an important question emerges: how can we use head impact sensors if half of the exposure data is discarded because of measurement issues?

Limitations

This study has several limitations in its design. We opted to use the mouthguard's data as the reference data to evaluate the patches based on previous research showing the superior coupling of the mouthguard to the skull.[219, 418] However, we also saw some level of decoupling in the mouthguard, and the lack of a true reference is an important limitation. Furthermore, there has been no formal evaluation of the validity of the sensors and impact exposures we used in this study. Other models of the Prevent Biometrics mouthguards have been shown to be well correlated with reference measurements and accurate in both helmeted and unhelmeted testing.[198, 232] To the best of our knowledge, the CSx patches have not been assessed in the laboratory, although the CSx mouthguard, which features the same electronics and algorithms, proved to be on par with other devices in terms of accuracy, or even better for angular kinematics (see Chapter 3). In the context of this PhD thesis, a laboratory study was originally planned in Vancouver, Canada, to assess the validity of the sensors used in the present study, but was forced to be cancelled because of the COVID-19 pandemic.

Several steps were taken in this study to ensure we compared acceleration events across sensors that were associated with the same specific head impact. However, it is impossible to know if the sensors were triggered at the exact same time. The temporal alignment of the sensors' traces in Figure 5.12 assumes so, but may be erroneous, especially given previous reports of skin patch motion lagging behind mouthguard motion,[418] and our observation of headgear decoupling. The misalignment would not affect the peak kinematics, but this is only valid if the sensors were triggered within 40 ms of each other (i.e., the duration of the post-trigger recording). If the sensors were triggered with a delay longer than 40 ms, which we cannot verify because of the 1-second resolution of the patches' timestamp, then the recordings may not be directly related. A patch being triggered more than 40 ms before or after the mouthguard could be a consequence of the sensor measuring motion artefacts crossing the trigger threshold. Cases where this happens could involve the participant bringing their gloves up to the headgear to protect themselves (triggering the patches) before being punched (triggering the mouthguard). A longer recording duration and an accurate high-resolution clock in the patches may have revealed a lag between the signals.

Also, we were not aware of the sensors' refractory time, i.e., how quickly is a sensor ready to capture a new impact. The shortest time between two subsequent recordings in our mouthguard data was 85 ms. While 85 ms is short compared to the 300-ms refractory time reported for another instrumented mouthguard,[197] it may compromise the data verification process when several impacts occur within a short window of time. It could also miss important impacts while the sensor is getting ready for the next impact.

It is known that noise in the angular velocity signals is amplified in the angular acceleration signals by the differentiation process. This is further projected onto the linear acceleration signals during the transformation to the head's CG (Equation 5.1).[219, 232, 418] From our examination of the raw and processed patch data, we were able to identify that the shape and magnitude of the LA signals at the head's CG were largely driven by the introduction of the angular acceleration cross-product term during the transformation (Figure 5.3). We reason this to be one of the main causes for the high PLA values observed in the 'bad' recordings. Wu et al. also reported five-to-seven-fold increases in the PLA differences between the mouthguard and the skin and skull cap sensors when the latter's linear acceleration was transformed to the head's CG, when compared with when the mouthguard's data was projected to these sensors' location.[418] To verify that a proportion of the differences between the mouthguard and the patches is explained by errors introduced by the transformation to the head's CG, we could have transformed the mouthguard's raw data to the patches' location. Nonetheless, there would still be several unknown steps in the mouthguard's processing algorithm that could influence the differences between the mouthguard's and patches' measurements.

While it has not been described in the literature, we were informed that the xPatch algorithm also filters the angular acceleration curves before using them for the transformation of the LA signals (communication between GS and the xPatch manufacturer). This may explain why the average PLA measured by the skin patch in this study were larger than previously reported by the xPatch (mean of 48.81 ± 26.68 g in this study vs. means comprised between 15 and 25 g in other studies in wrestling, taekwondo or soccer[204, 302, 335]). These differences in PLA highlight the large influence that seemingly small differences in processing methods, often unknown to the end-user, can have on the results. Thus, the comparison of results from studies using different devices and processing algorithms is challenging. Future research would benefit from increased transparency from the manufacturers and from the standardisation of these processes.

To calculate the patches linear acceleration at the head's centre of gravity, we used custom transforms with estimates of the location and orientation of the patches for each participant. Most commercial systems use generic transformation parameters (e.g., the Prevent Biometrics algorithm is based on a 50th percentile male NOCSAE headform), but there has been a report of custom transforms being used by a research team.[339] Transforming the linear acceleration to the head's CG allows the comparison with other studies, and we presume that customizing the transformation parameters improves the accuracy of the linear acceleration.[219] However, the influence of errors in estimating these parameters on the final traces may be significant, particularly the estimation of the axes orientation.[219, 221] Of note, we did not account for parallax or image distortion when estimating the patches' location and orientation, and the position of landmarks, particularly the occiput, was highly uncertain in several cases (Figure 5.1). Consequently, we do not know if the transformed data are more or less accurate with our custom transformations than if we had used a generic transformation. This may ultimately have affected the association between the patches and mouthguard data.

Additional limitations pertain to the quality assessment of the sensor's raw data. First, the analysis of 'good', 'questionable' and 'bad' recordings was conducted on matched recordings, and excluding singled-out recordings significantly affected the summary metrics for the patches (Figure E.1, Figure E.2, Figure E.3, in Appendix E). As a result, the proportions of 'good' recordings reported may not be representative of the whole dataset. Second, we observed a higher proportion of 'questionable' recordings in the headgear patch dataset compared to the other two sensors. The dichotomy between good and bad was not as clear in many cases, and the application of the criteria more complex (e.g., we observed a signal inversion with raw LA values from -20 to +20 g, happening over 20-30 ms). In most of these cases, we could not determine if the motion measured could be associated with skull motion or independent headgear motion. Looking deeper into head and headgear motion for impacts between a padded glove and a padded headgear, in a controlled setting with high-speed cameras, for example, may be helpful in better understanding the influence of the headgear on the kinematics measured by the patch. Third, it is important to note that the application of the criteria was subjective. As such, the "sharpness" of the pulses was ultimately left to the appreciation of the rater and the classification may have differed with another rater. Finally, our approach was conservative, and the rater tended to classify recordings as 'questionable', and therefore exclude them if there was any doubt as to the quality of the signal. Researchers planning to conduct quality assessments of the data may benefit from a set of quantitatively defined criteria.

5.7. Conclusion

In summary, our analysis of 442 video-verified head impacts explored the quality of kinematic signals from matched acceleration events from mouthguards, skin patches and headgear patches. It highlighted large proportions of recordings containing measurement issues, suggesting that the kinematic data, particularly from the patches, often reflects the motion of the sensor itself rather than the motion of the head. These issues were more frequent when the impact occurred in the proximity of the sensor and led to over-estimations of the peak accelerations. There was little to no association in peak kinematics between the patches and the mouthguard. As a result, the kinematics measured by the skin and headgear patches should not be used for the analysis of boxing-related exposure to head impacts as their magnitude is affected by the patches' poor coupling to the skull. While the mouthguard showed a smaller proportion of poor-quality recordings, we recommend that processing of its data include both video verification and quality assessment. Research is required to improve the coupling of the mouthguard to the skull and determine how to use exposure datasets where large proportions of data are missing because of measurement issues.

Chapter 6 - Individual-specific associations between head impact kinematics and acute symptoms changes resulting from boxing sparring: An exploratory study

This manuscript is currently in preparation for journal submission.

Elements of protocol and supplementary materials can be found in Appendix D and Appendix E, respectively.

6.1. Prelude

The previous chapters have found that video verification and quality assessment of head acceleration recordings is essential to robustly quantify the exposure to head impacts. Despite a large proportion of recordings suggesting skull/sensor decoupling, the mouthguard proved to be the best solution, out of the sensors tested, to evaluate the severity of the exposure. We therefore utilised video-verified, valid mouthguard data to explore the kinematics of the head during boxing sparring. Specifically, we investigated the occurrence of boxers' self-report of acute concussion symptoms and their association with head kinematics, via the appraisal of individual-specific approaches.

6.2. Abstract

Background: Because of the large range of acceleration magnitude reported for concussive events, the acute dose-response relationship between head impact exposure and the development of concussion remains elusive. Previous research has suggested that individual-specific approaches and the accumulative effects of head impacts should be investigated.

Objectives: To monitor acute changes in concussion-related symptoms resulting from boxing sparring, and assess how these symptoms evolve over the subsequent 48 hours, and how they relate to head impact exposure.

Methods: Seven competitive boxers participated in this feasibility study. Symptoms of concussion were captured via the SCAT5 symptom scale upon recruitment, before, immediately after, and 48 hours after one to three sparring sessions for each participant. Head impact data was collected with instrumented mouthguards and included after video-verification and quality assessment. Each session's highest-magnitude impact and cumulative burden for several injury severity metrics were qualitatively analysed with respect to acute changes in symptoms score.

Results: Group-based analyses did not show any difference in symptoms score pre- and post-sparring, but three participants reported changes in symptoms in four instances. In most of these

cases, the symptom severity increased from pre- to post-sparring, then decreased to baseline levels after 48 hours. There was no strong association between session-specific cumulative burden and change in symptoms score.

Discussion: Symptoms may have resulted from the combination of several factors, namely pre-existing dispositions, several above-average-magnitude impacts, and/or a high density of head impacts. The existence of a risk group requires further research, as it may have implications for the identification of injurious impacts and the management of these athletes' training and recovery. The small sample limited the strength of the findings.

Conclusions: Our results support the concept of individual-specific analyses, as this approach allowed us to identify patterns emerging from a few athletes, that was not visible in cohort-averaged summaries.

6.3. Introduction

During boxing sparring, commonly performed multiple times a week,[126, 159] competitive boxers can sustain between 3 and 8 head impacts per minute.[54, 378] This rate is higher than in American football (0.1 to 0.5 impacts per minute in games or full-contact drills)[10, 60, 120] and the average peak linear acceleration (PLA) measured by head impact sensors during boxing sparring is as high as in other contact sports (22-30 g).[54, 302, 378] The number and intensity of head strikes sustained as part of normal training may lead to the presence of symptoms and therefore to an underlying insult to the brain. The presence of symptoms can in turn affect both the immediate performance and the risk of subsequent injury. While previous studies have suggested that participating in amateur boxing may be insufficient to cause chronic neurologic disorders,[163] there is little information about how participation in sparring affects an athlete's neurologic and cognitive status during the training week. Neuropsychological and cognitive testing conducted on boxers before and after a sparring session showed impairments in delayed and verbal memory[54, 226, 378] and in reaction time.[54] The most common self-reported symptoms after a concussive event in combat sports were headaches, fatigue and dizziness,[159] but concussion-related symptoms resulting from sparring have not been assessed.

It is common for combat sports athletes to not seek medical attention after a concussion, and to resume full-contact training shortly after injury.[30, 126, 159, 238] These behaviours are primarily thought to be due to a lack of understanding of concussion symptoms and their consequences,[30, 126] and to the athletes' unwillingness to report concussion symptoms.[30, 238] Therefore, as in other sports, there is a need for tools that would allow an objective and accurate monitoring of an athlete's state with respect to brain injury,[42] enabling informed training and recovery practices and aiming at improving athlete's performance and health in the short-to-long term. Particularly, with the use of wearable head impact sensors, we can investigate the possible dose-response relationship between head impacts kinematics and the signs and symptoms of concussion. If such a relationship exists, it could be a valuable tool to manage an athlete's impact load by, for example, proactively stopping a sparring session before reaching a level of head impacts exposure that is associated with the athlete experiencing symptoms.

The acute dose-response relationship between head impact kinematics and the signs and symptoms of concussion has been investigated for more than 15 years, but because of the large range of acceleration magnitudes and directions reported for concussive events,[38, 149] a global injury threshold remains elusive.[269] Additionally, associations between symptoms or neurocognitive performance and impact exposure could not be found.[41, 149, 352, 378, 412] Several factors may contribute to the lack of association, the first being issues related to the head impact sensors. High numbers of acceleration events representing the motion of the sensor independent from the skull, along with their associated larger peak accelerations (see Chapters 4 and 5), may have contributed to obscuring important associations. Furthermore, the validity of the kinematics measurements for individual impacts is poor for several types of sensors[367, 369, 418] and particularly for angular kinematics, which are especially relevant to injury.[182, 212]

There have been multiple reports that impacts that could be identified as concussive were the highest magnitude, or amongst the highest magnitude, impacts recorded for an individual.[120, 134, 352, 374] This suggests that tolerance to head impacts is dependent on intrinsic factors and that individual-specific approaches should be investigated.[342, 352] It was also suggested that repetitive head impacts decrease an athlete's tolerance,[374] which refers to the minimum head impact magnitude needed for the athlete to present signs or symptoms of concussion. While it is possible to link a specific head impact to a concussion if an athlete immediately displays signs of altered state,[101] athletes can also experience delayed concussion-related symptoms that cannot be linked to one specific impact event.[26, 27, 41, 409] It has been shown that athletes whose symptoms' onset could not be identified, and thus had a delayed diagnosis of concussion, sustained twice as many head impacts on days with injury than on days without injury.[27] Studies have measured acute changes in concussion-related symptoms,[412] neurocognitive performance,[378] and blood biomarkers[188, 191, 192, 353, 424] following a single contact-sport session even without a concussion diagnosis. Collectively, these studies show that concussion, or the onset of concussion-related signs and symptoms, may result from a combination of multiple sub-concussive head impacts and/or moderate-to-high magnitude impacts.

The goal of this observational study was to assess the feasibility of monitoring acute changes in concussion-related symptoms in boxers after sparring, and how these symptoms evolve over the subsequent 48 hours. The study also used an individual-specific approach to explore head impacts exposure over several sparring sessions, focusing on video-verified good-quality mouthguard measurements. As the sample size was too small to formally test the hypothesis that individual-specific exposure was associated with acute changes in self-reported symptoms, we qualitatively examined whether session-related cumulative exposure or highest-severity impacts could be linked to such changes. We also tested the feasibility of using control charts to monitor exposure using individual-specific thresholds.

6.4. Methods

Overview

This chapter explores acute self-reported symptoms of concussion and head impact exposure data resulting from boxing sparring. Data were collected over one to three sparring sessions each for seven competitive boxers (4 females, 3 males). Symptoms of concussion were captured via questionnaires at various time points: upon participant recruitment, and before, immediately after, and 48 hours after each sparring session. In this chapter, head impact exposure is quantified from video-verified head acceleration events that were recorded with instrumented mouthguards and whose kinematic data met raw signal quality criteria. Collection and processing methods for head impact data are fully described in Chapters 4 and 5.

Participants are identified via the number of their mouthguard (MG03 to MG10). MG06 withdrew before the start of the study due to injury, and MG01 and MG02 were kept as back-up units and were not used.

Data collection and processing

A total of 16 athlete-sessions were observed, with one to three athlete-sessions per participant. An athlete-session was complete if head impacts were measured by instrumented mouthguards and could be verified by video for the entirety of the sparring session. Five incomplete athlete-sessions were excluded because of camera malfunction or the athlete sparring outside of the area covered by the cameras.

Self-reported concussion history and symptoms

Each participant completed a questionnaire at the beginning of the study about their experience of combat sports, training habits and concussion history (Appendix D, page 217). Specifically, the participants were asked how many times they had been knocked out and/or lost consciousness or been declared "*unfit to continue fighting or sparring as a result of a blow or series of blows to the head*" (by themselves, a referee, a trainer or a ringside physician), and how often these stopped fights/sparring sessions had resulted in a diagnosed concussion.[85] Participants also reported whether they had been diagnosed with a concussion from any other activity over their lifetime. A definition of concussion[6, 341] was then provided and participants were asked to recall how many concussions they thought to have sustained while fighting or sparring over their lifetime.

> *"Some people have the misconception that concussions only happen when you black out after a hit to the head or when the symptoms last for a while. In reality, a concussion has occurred anytime you have had a blow to the head that caused you to have symptoms for any amount of time. These include: blurred or double vision, seeing stars, sensitivity to light or noise, headache, dizziness or balance problems, nausea, vomiting, trouble sleeping, fatigue, confusion, difficulty remembering, difficulty concentrating, or loss of consciousness. Whenever anyone is being "rocked", "wobbled" or "fazed", that too is a concussion."*

Next, participants were questioned about training habits[30, 185] and sparring-related strikes to the head: how often do they receive strikes to the head of extremely light, light, moderate or heavy intensity,[126] and how often do they take "*at least one heavy blow to the head that "rocks" or "fazes" them, or leaves them feeling dizzy, off-balanced or with blurry vision, even for a short time?*" Subsequently, the list of 22 concussion-related symptoms from the Sport Concussion Assessment Tool, 5th Edition (SCAT5),[159] was presented and participants reported how often they sustain each symptom within 24 hours of a sparring session (never, sometimes, about half the time, most of the time or always). Finally, participants were asked if they had a history of attention deficit or hyperactivity disorder (ADHD), learning disorder and/or migraines.[379, 425]

Participants also completed short questionnaires (Appendix D, page 221) around each sparring session, which took place on Saturday mornings (except for one Wednesday evening session): before their warm-up (PRE, mean: 30 ± 10 minutes before the first sparring round, range: 11-45), after their cool-down (POST, mean: 32 ± 24 minutes after the end of the last round, range: 10-111), and approximately 48 hours after (mean 53 ± 7 hours, range 47-69). Each questionnaire asked for the intensity (0 – None, to 6 – Severe) for the 21 or 22 symptoms of the SCAT5 ("Trouble falling asleep" was excluded at PRE and POST), as well as the participant's overall feeling ("*If 100% is feeling perfectly normal, what percent of normal do you feel?*"). The 48-hour questionnaire also asked the participants to recall how they felt at their worst, defined as the time when they experienced symptoms at the highest intensity (WOR). We therefore collected self-reported symptoms at four time points: PRE, POST, WOR, 48H. Additionally, the POST questionnaire asked the participants if they had sustained heavy blows, along with a description to identify them on video. The participants completed all questionnaires with no intervention from the researcher, either on a tablet at the gym or on their own device outside of their training time.

The total number of self-reported symptoms was calculated (range 0-21 or 0-22), as well as the total symptoms severity score (TSSS), computed as the sum of every symptom's severity (each 0-6, for a TSSS ranging from 0 to 126 or 132). Symptoms were grouped into clinical profiles: oculo-vestibular, cognitive-fatigue, post-traumatic migraine and anxiety/mood (Table 6.1), except for neck pain and sleep-related issues as per their role as modifiers in concussion assessments.[139, 216] These clinical profiles were used to determine if there was a pre-dominant profile emerging from participation in sparring. We calculated the PRE-POST, PRE-WOR and PRE-48H changes in the number of symptoms, TSSS, and feeling of normal. By doing this rather than comparing post-sparring scores to a baseline report or normative data, we aimed to normalise the scores by accounting for pre-existing symptoms on the day of data collection, thus removing both within- and between-subject variability.[329]

Table 6.1. Clusters of symptoms based on clinical profiles of sport-related concussion.

Clinical profile	Symptoms
Oculo-vestibular	Dizziness, Blurred vision, Balance problems, Nausea or vomiting
Cognitive-fatigue	Feeling slowed down, Feeling like "in a fog", "Don't feel right", Difficulty concentrating, Difficulty remembering, Fatigue or low energy, Confusion, Drowsiness
Post-traumatic migraine	Headache, "Pressure in head", Sensitivity to light, Sensitivity to noise
Anxiety/mood	More emotional, Irritability, Sadness, Nervous or Anxious

Kinematics

Participants were equipped with Prevent Biometrics Hybrid mouthguards (Prevent Biometrics Inc., Edina, MN). They are boil-and-bite mouthpieces that incorporate a triaxial linear accelerometer (ADXL372, Analog Devices, Boston MA, range ± 200 g) and a triaxial angular rate sensor (BMG250, Bosch, Gerlingen Germany, range ± 35 rad.s^{-1}), both sampling at 3200 Hz. The sensors were programmed to record a 50 ms acceleration event (10 ms pre- and 40 ms post-trigger) any time the linear acceleration signal reaches 10 g on any axis. Prevent Biometrics provided us with the raw 3-degrees of freedom (3DOF) time-series data from the accelerometer and angular rate sensor for all acceleration events. Processed 3DOF time series data were downloaded for the linear acceleration (LA), angular velocity (AV) and angular acceleration (AA) from the Prevent Biometrics user portal. These processed data were filtered and transformed to the head's centre of gravity by Prevent Biometrics' proprietary algorithms.

Sparring was filmed using a minimum of two cameras covering the sparring area, set up with a frame rate of 60 fps, a shutter speed of 1/120 s, and a resolution of 1080 p. A mouthguard calibration impact was performed at the start of the session in the view of one camera to align the mouthguards' timestamps to the video timeline. All other cameras were later synchronised to this first camera. The videos of all rounds for each complete athlete-session were reviewed and coded with Nacsport Elite 6.0.0 (Nacsport, Canary Islands, Spain) by one rater (ELF) to identify head and body impacts, following the methods outlined in Chapter 4.

Data reduction and metrics calculation

First, the acceleration events recorded outside of sparring time were excluded. Second, the remaining events were matched to video events, independent of the Prevent Biometrics classification algorithm. Finally, only the video-verified recordings satisfying the raw signal quality criteria established in Chapter 5 (Table 5.1, page 86) were included for further analysis. We included all such events, independent of their magnitude. All subsequent steps were performed on MatLab (R2019a, MathWorks, Natick, MA). The following ten kinematic metrics were calculated from the processed 3DOF time series data for every included acceleration event.

Peak linear acceleration, angular velocity, and angular acceleration (PLA, PAV, PAA) were defined as the maximum value of their respective resultant time series. Peak accelerations are the most commonly reported kinematic metrics in head impact research (see Chapter 2). Video-verified mouthguard measurements in collegiate American football and combat sports have shown average and median non-concussive PLA and PAA values comprised between 18 and 37 g and 1240 and 3355 rad.s^{-2}, respectively.[164, 220, 389, 416] Other studies on American football have used non-concussive and concussive impacts to propose injury risk thresholds, with 25%, 50% and 80% risks of sustaining a concussion being associated with PLA and PAA values of 66, 82, 106 g and 4600, 5900, 7900 rad.s^{-2}, respectively.[423] More recently, concussive impacts in high school and collegiate American football athletes were summarised with average PLA and PAA values of 98.68 g [95% CI 82.36 – 115.00] and 5776.60 rad.s^{-2} [4583.53 – 6969.67].[38]

The maximal change in angular velocity ($\Delta\omega_{\max}$) was calculated as the largest excursion from the angular velocity at the onset of impact.

$$\Delta\omega_{max} = \max\left(\sqrt{\sum(\omega_i - \omega_{i0})^2}\right) \quad \textit{Equation 6.1}$$

Where ω_i is the angular velocity over time and ω_{i0} the velocity at the onset of impact, measured about the x-, y- and z-axes. The onset of impact corresponded to the sample that triggered the recording, i.e., when any axis of the raw LA reached ±10 g. The maximal change in angular velocity has been proposed as a predictor for head injury and brain strains.[212, 213, 243] It is rarely used in sport-related head impact research, but authors have reported average values around 10 to 14 rad.s^{-1}.[164, 416] Two concussive events have been recorded using mouthguards at 23 and 34 rad.s^{-1}.[164]

The Gadd Severity Index (GSI) is a linear-acceleration-based injury criterion originating from the Wayne State Tolerance Curve (WSTC).[137] In the present study, it was calculated over the first portion of the signal for which the resultant linear acceleration was over 4 g.[164]

$$GSI = \int a(t)^{2.5} dt \qquad \text{Equation 6.2}$$

Where $a(t)$ is the resultant linear acceleration. The WSTC and the GSI were developed from animal and cadaver research and used to quantify severe brain injury. A head acceleration resulting in a GSI value over 1000 was associated with a serious internal head injury and danger to life.[137] Later, a value of 300 was proposed as a tolerance threshold for concussion[311] and concussive impacts in American football have been reported with median or mean GSI values of approximately 250.[27, 229] The GSI is seldomly used in head impact research, but average non-concussive values have been reported between 26 and 66.[164, 290, 378]

The Head Injury Criterion (HIC) was later adopted as a modification of the GSI, focusing on the main linear acceleration pulse.[402]

$$HIC = \left[\frac{1}{t_2 - t_1} \int_{t_1}^{t_2} a(t) dt\right]^{2.5} (t_2 - t_1) \qquad \text{Equation 6.3}$$

Where $a(t)$ is the resultant linear acceleration, and t_1 and t_2 are two time points chosen to maximise the value of HIC. In the present study, we report the HIC_{36}, where $(t_2 - t_1)$ is bounded to ≤ 36 ms. Apart from PLA and PAA, the HIC is the most commonly reported injury severity metric in head impact research. A critical value of 250 has been proposed as a concussion tolerance level,[311] and concussive impacts in American football have been previously measured with a median HIC value of 178.8,[229] or a mean HIC value of 345 ± 270.[27] Only one study summarising video-verified mouthguard recordings reported HIC values for non-concussed football players (median of approximately 35).[164] Other studies on various sports and using various technologies reported average non-concussive values between 13 and 43.[290, 302, 378]

The Rotational Injury Criterion (RIC) is the angular equivalent of the HIC and has also been proposed as a predictor for brain strains.[200]

$$RIC = \left[\frac{1}{t_2 - t_1}\int_{t_1}^{t_2} \alpha(t)dt\right]^{2.5} (t_2 - t_1)$$ *Equation 6.4*

Where $\alpha(t)$ is the resultant angular acceleration, and t_1 and t_2 are two time points chosen to maximise the value of RIC. In the present study, we report the RIC_{36}, where $(t_2 - t_1)$ is bounded to ≤ 36 ms. A critical RIC_{36} value of 1.03E+07 has been offered to represent a 50% risk of concussion.[200] Only one study on sports-related head impacts reported RIC values, with non-concussive impacts resulting in a median RIC value of approximately 2E+06, and two concussive impacts at approximately 1.2E+07 and 9E+07.[164]

The Brain Injury Criterion (BrIC) is a metric based on 3DOF angular velocity, developed from numerical studies and proposed as another predictor for brain strains.[385]

$$BrIC = \sqrt{\sum\left(\frac{\max\left(|\omega_i - \omega_{i0}|\right)}{\omega_{iC}}\right)^2}$$ *Equation 6.5*

Where ω_i are the angular velocity over time, ω_{i0} the velocity at the onset of impact measured about the x-, y-, and z-axes. We used the angular velocity changes from the onset of impact rather than the absolute maximum velocities to account for the fact that BrIC was developed using head impacts where the initial angular velocity was at or near zero. The onset of impact corresponded to the sample that triggered the recording. ω_{iC} are critical angular velocities and are equal to 66.3, 53.8, and 41.5 $rad.s^{-1}$ about the x-, y-, and z-axes, respectively.[385] Values for the BrIC are estimated to range between 0 and 3, with a higher number representing a higher risk of injury. A BrIC of one is nominally equal to a 50% risk of AIS 4+ anatomic brain injury.[385] Only one study on sports-related head impacts reported BrIC values, with non-concussive impacts resulting in a median BrIC value of approximately 0.1, and concussive impacts between 0.4 and 0.5.[164]

The generalized model for brain injury threshold (GAMBIT) was developed to consider the combined effects of linear and angular accelerations on injury risks.[293] The simplified equation for GAMBIT was used in the current study:

$$GAMBIT = \frac{a_{peak}}{250} + \frac{\alpha_{peak}}{10{,}000}$$ *Equation 6.6*

Where a_{peak} and α_{peak} are the peak linear acceleration and peak angular acceleration, respectively. The 250 g and 10,000 $rad.s^{-2}$ denominators represent the critical values chosen as tolerance thresholds. Based on these thresholds, a GAMBIT value higher than 1 is associated with a brain injury. The only study reporting GAMBIT values from sport-related head impacts measured non-concussive impacts with a median value of approximately 0.17 and concussive impacts at approximately 0.43 and 0.63.[164]

The Head Impact Power (HIP) was developed based on laboratory reconstructions of American football impacts.[294] It is an expression of the head's rate of change of linear and angular kinetic energy and is defined by the equation below.

$$HIP = \max\left(ma_x(t)\int a_x(t)dt + ma_y(t)\int a_y(t)dt + ma_z(t)\int a_z(t)dt + I_{xx}\alpha_x(t)\int \alpha_x(t)dt + I_{yy}\alpha_y(t)\int \alpha_y(t)dt + I_{zz}\alpha_z(t)\int \alpha_z(t)dt\right) \quad \text{Equation 6.7}$$

Where m is the mass of the head fixed at 4.5 kg; a_x, a_y, and a_z are the linear accelerations (in m.s^{-2}) along the x-, y-, and z-axes, respectively; I_{xx}, I_{yy}, I_{zz} are the head's moments of inertia about the x-, y-, and z-axes, fixed at 0.016, 0.024 and 0.022 N.m.s^2, respectively; and α_x, α_y, α_z are the angular accelerations about the x-, y-, and z-axes, respectively. The linear and angular accelerations were integrated from the onset of impact (the sample that triggered the recording, to assume initial velocities of 0 m.s^{-1} and 0 rad.s^{-1}) to the end of the recording, i.e., over 40 ms. The HIP provides a value in Watts (W), and a critical value of 12.8 kW was associated with a 50% probability of concussion.[294] Since then, two sport-related head impact studies reported HIP values, with non-concussive impacts resulting in average HIP values of 2.8 to 7.15 kW, and concussive impacts in HIP values between 10 and 28 kW.[164, 389]

Analysis

To explore the occurrence of symptoms, the participants' history of concussion and sparring-related symptoms are reported for each participant and then summarized over the cohort with means and standard deviations (SD). The number of symptoms and TSSS at POST, WOR and 48H are compared to PRE using Friedman's test and completed with post-hoc tests with a Dunn-Sidák correction to assess the differences between time points. The significance level was set at 0.05. Additionally, the evolution of the number of symptoms and TSSS over the four time points are plotted for each athlete-session. Qualitative analysis is provided for the athlete-sessions where an increase in TSSS ≥ 5 was reported between PRE and any of the post-sparring time points.

To investigate the association between head impact kinematics and symptoms, the mean and SD, median and inter-quartile range (IQR), maximum value and cumulative burden (the sum of the magnitude of all acceleration events included for an athlete-session) were calculated by athlete-session for each of the ten kinematic metrics, then summarized for the full cohort. The association between acute changes in self-reported symptoms and cumulative burden or maximum value is narratively described for the two participants who reported a change in TSSS ≥ 5 and for which several sessions were collected over the study. For each metric, the mean, median, maximum value and cumulative burden of each athlete-session were also normalised over the participant's mean value calculated over two or three athlete-sessions. The normalised data were plotted to explore the collinearity between the metrics.

Ten control charts (one for each metric) were established for each athlete-session. While control charts traditionally represent count data to monitor the incidence of illness or injury,[230, 362] we use them to show the magnitude of all acceleration events over time and highlight outliers. For each participant, we calculated the mean and SD of all events sustained over all available athlete-sessions and plotted the participant's mean as well as two upper-control limits, one and two standard deviations above the mean.[278] All acceleration events with a magnitude over the upper-control limits were identified and we specifically highlighted the events exceeding these limits consistently across all ten metrics (i.e., outliers). Such events may be particularly relevant as they reflect a higher-than-average severity on metrics that represent different aspects of head motion (e.g., linear vs. angular, resultant vs. individual components). We qualitatively analysed such events relative to the presence or absence of self-reported symptoms.

6.5. Results

Self-reported history

From the questionnaire completed upon recruitment, none of the seven participants had a history of loss of consciousness or knock-out due to blows to the head while fighting or sparring (Table 6.2). Two participants reported having been declared unfit to continue fighting/sparring at least once but did not seek medical attention. After being given the definition of concussion, five participants reported having sustained at least one concussion (mean of 3.4 ± 4.0 estimated concussions per athlete). The participants typically sparred at intensities ranging from moderate to as intense as competition, with strikes to the head being of moderate intensity on average. Four participants reported more than 10 symptoms being sometimes experienced within 24 hours of sparring, while the other participants generally experienced few or no symptoms. The most common symptoms reported were "Headache", "Neck pain", "Fatigue or low energy", and "Trouble falling asleep", with the latter three sometimes persisting for longer than 24 hours. One participant (MG07) reported a history of ADHD; no other participant reported any co-morbidities.

Table 6.2. Participants' details, self-reported history of concussion, and frequency of symptoms experienced within 24 hours of prior sparring sessions, collected via questionnaire upon recruitment.

Part.	Sex	Age	Years of boxing	KO or LOC*	Declared unfit*	Diagnosed concussions*	Estimated concussions*	Frequency of heavy blows	Number of symptoms (scale of 0 to 22) "Sometimes"	"About half the time"
MG03	F	22	5	0/0	1/3	0/0	4/4	Sometimes	11	3
MG04	F	29	3	0/0	0/0	0/0	0/3	Sometimes	11	0
MG05	F	22	5	0/0	0/0	0/0	0/0	Never	6	0
MG07	F	23	1	0/0	0/0	0/0	0/1	Sometimes	13	1
MG08	M	20	2	-/0	-/1	-/0	-/>10	Sometimes	18	1
MG09	M	23	8	0/0	0/0	0/0	1/1	Never	1	0
MG10	M	24	6	0/0	0/0	0/0	0/0	Never	0	0
Mean ± SD		23.3 ±2.8	4.3 ±2.4	0/0	0.1 ±0.4 / 0.6 ±1.1	0/0	0.7 ±1.5 / 2.7 ±3.5		8.6 ±6.6	0.7 ±1.1

F: Female, M: Male, KO: knock-out, LOC: loss of consciousness
*The numbers of KO, LOC, unfit decision, and concussions are reported as: from fighting / from sparring. The participant MG08 had no fighting experience, therefore only sparring-related numbers are reported.

Sparring-related symptoms

Sixteen full athlete-sessions were collected, with an average number of 9 ± 2 rounds performed per session (range: 6 to 12), and an average time spent sparring of 27 ± 7 minutes (range: 18 to 39).

Overall, the number of symptoms and TSSS were significantly lower at 48H than at PRE (number: $F_r(3) = 10.67$, $p = 0.021$; TSSS: $F_r(3) = 9.07$, $p = 0.028$)(Table 6.3). The feeling of normal was also significantly higher at 48H than at POST ($F_r(3) = 10.15$, $p = 0.017$). No other difference was found. The feeling of normal was sometimes dissociated from the symptoms, with decreases larger than 10% (up to 40%) despite the absence of symptoms, or an increase larger than 10% despite a rise in symptoms (Table 6.3).

Three participants reported experiencing an increase in TSSS ≥ 5 between PRE and any post-sparring time point in a total of four occasions (25% of all athlete-sessions)(Table 6.3, Figure 6.1, Figure E.4). In three of the four cases, we observed a typical exposure-recovery curve, where the TSSS increased from PRE to POST (mean increase of 14.3 ± 2.3), then decreased from POST to WOR (-9.7 ± 8.6) and from WOR to 48H (-8.7 ± 1.2)(Figure 6.1). In the fourth case, the TSSS decreased immediately after sparring (from 7 to 1 at POST) then increased a few hours later (1 to 12 at WOR) before returning to baseline (12 to 7 at 48H). The cognitive-fatigue symptoms cluster was affected in all four instances, while the oculo-vestibular cluster showed no or little change over the 48-hour time window (Figure 6.1). The time where the participants felt at their worst following sparring, in the five instances where symptoms were reported in the 48-hour questionnaire, was on average 5 ± 1.5 hours after the end of the sparring session (range 2.7 – 6.5).

Table 6.3. Evolution of the number of symptoms, total severity scores, and feeling of normal for each athlete-session.

Athlete-session	Exposure (minutes)	Number of symptoms				Total symptoms severity score				Feeling of normal (%)			
		PRE	POST	WOR	48H	PRE	POST	WOR	48H	PRE	POST	WOR	48H
MG03													
Session 1	30	1	0	0	0	1	0	0	0	95	80 ↓	100	100
MG04													
Session 1*	30	9	13 ↑	4 ↓	1 ↓	16	29 ↑	10 ↓	2 ↓	60	75 ↑	70 ↑	90 ↑
Session 2	27	3	4	2	1	5	4	4	1	95	90	90	98
Session 3	38.5	5	3	0 ↓	0 ↓	7	3	0 ↓	0 ↓	95	98	100	100
MG05													
Session 1	33	1	1	0	0	2	1	0	0	90	90	100	100
Session 2	24	0	0	0	0	0	0	0	0	100	100	100	100
Session 3	18	0	0	0	0	0	0	0	0	100	100	100	100
MG07													
Session 1	30	1	0	0	0	1	0	0	0	97	96	100	100
Session 2*	18	3	13 ↑	7 ↑	4	5	22 ↑	14 ↑	6	93	88	85	90
Session 3*	30	6	1 ↓	5	5	7	1 ↓	12 ↑	7	90	95	85	95
MG08													
Session 1*	27	1	8 ↑	6 ↑	2	1	14 ↑	12 ↑	2	90	90	40 ↓	90
MG09													
Session 1	24	0	0	0	0	0	0	0	0	100	100	100	100
Session 2	36	0	0	0	0	0	0	0	0	100	100	100	100
Session 3	21	0	0	0	0	0	0	0	0	100	100	100	100
MG10													
Session 1	18	1	0	0	0	1	0	0	0	100	80 ↓	100	100
Session 2	35	2	0	0	0	2	0	0	0	100	60 ↓	100	100
Mean ± SD	27 ± 7	2.1 ± 2.6	2.7 ± 4.6	1.5 ± 2.5	0.8 ± 1.6	3.0 ± 4.3	4.6 ± 9.0	3.3 ± 5.4	1.1 ± 2.2	94 ± 10	90 ± 11	92 ± 16	98 ± 4
Median [IQR]	28 [23 – 31]	1 [0 - 3]	0 [0 - 3]	0 [0 - 3]	0 [0 - 1]	1 [0 - 5]	0 [0 - 3]	0 [0 - 6]	0 [0 - 1]	96 [92 - 100]	93 [86 - 100]	100 [89 - 100]	100 [97 - 100]
Difference with PRE, p-value			0.97	0.16	0.02		0.85	0.34	0.02		0.93	1.00	0.16

PRE: pre-sparring, POST: immediately post-sparring, WOR: self-reported worst time where the participant experienced symptoms at the highest intensity, 48H: approximately 48 hours post-sparring, SD: standard deviation, IQR: inter-quartile range.

↑ / ↓ The arrows indicate an increase or decrease of POST, WOR or 48H relative to PRE by 3 symptoms, a total symptoms severity score of 5, or a feeling of normal of 10%, respectively. The difference between PRE and POST, WOR or 48H was assessed with a Friedman's test and the p-values reported were generated by post-hoc tests with Dunn-Sidák correction.

* indicates the sessions where the participant reported an increase in total symptom severity score equal or greater than 5 between pre-sparring and any time point post-sparring.

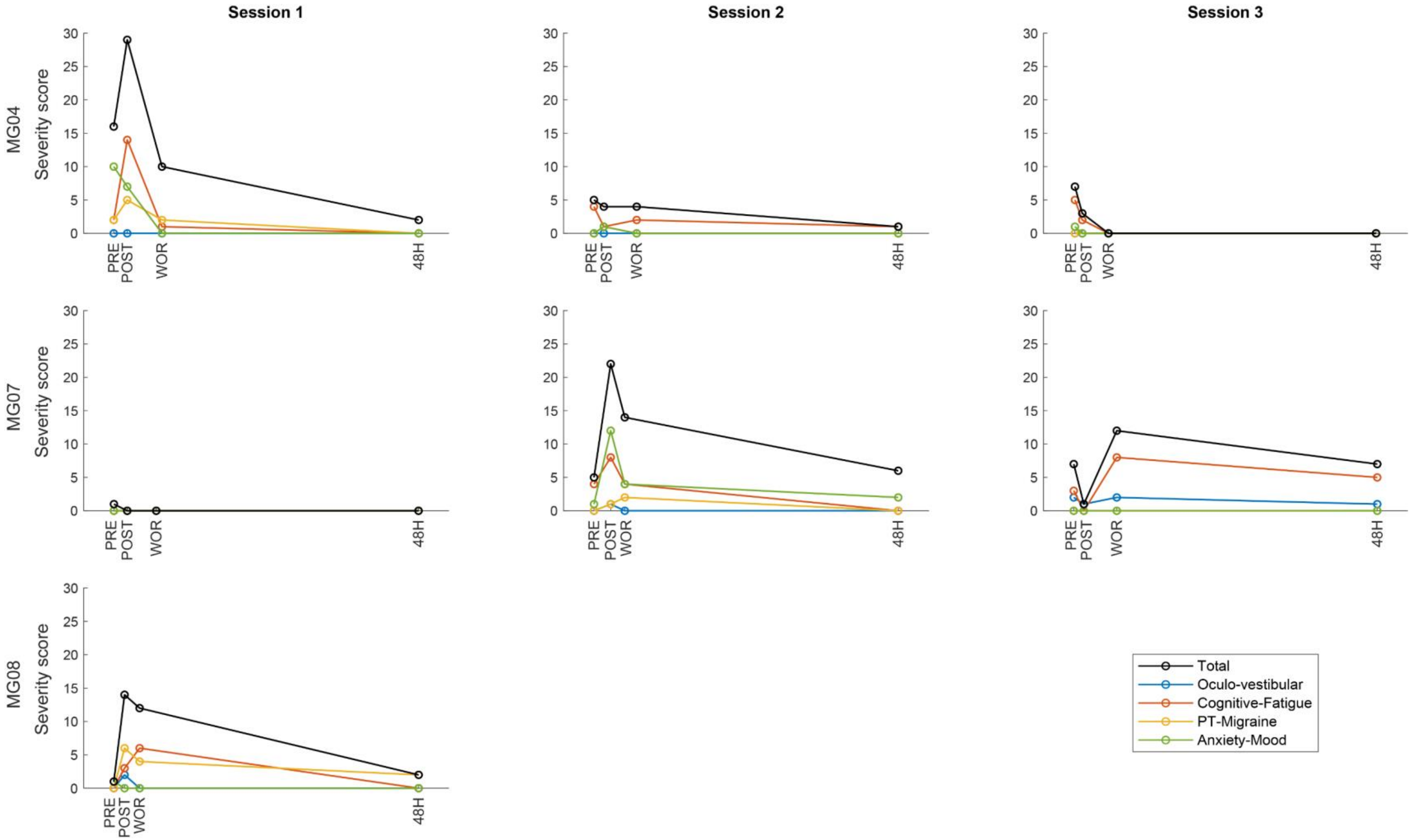


Figure 6.1. Evolution of the total symptom severity scores (TSSS) for the participants that reported a change in TSSS higher than 5 at least once over the course of the study. Only one session was collected from participant MG08 (bottom row). The TSSS (black line), as well as the severity scores by clinical profile (coloured lines), are represented for each athlete-session. The difference between the total and the sum of the clinical profiles scores is explained by the exclusion of the "Neck pain" and "Trouble falling asleep" symptoms from the clinical profiles. PRE: pre-sparring, POST: immediately post-sparring, WOR: self-reported worst time where the participant experienced symptoms at the highest intensity, 48H: approximately 48 hours post-sparring.

Kinematics

A total of 963 acceleration events were matched to video-verified head impacts. The overall proportion of recordings satisfying the quality criteria was 50% and ranged between athlete-sessions from 30 to 71%. There were variations between a same athlete's sessions, up to a 28%-difference. The final number of 'good' video-verified mouthguard recordings included in the analyses was 482. The number of such recordings ranged from 6 to 62 per athlete-session (mean: 30.1 ± 17.9 per session, or 1.2 ± 0.8 per minute of sparring)(Table 6.4).

Qualitatively, there were inconsistencies in the relationship between the change in TSSS and the sessions' cumulative burdens or maximum values for the two participants who reported acute changes in symptoms and for which more than one session were completed (Table 6.4, Table 6.5, Figure 6.2). The athlete-sessions with the most acceleration events and highest cumulative burdens (session 1 for MG04 and session 3 for MG07) were associated with the largest and second-largest increase in TSSS for these participants, respectively. However, the session with the largest increase in TSSS for MG07 (session 2) was associated with the lowest number of acceleration events, and the lowest cumulative burdens and maximum values for all metrics. Also, the session with the highest maximum values for MG04 (corresponding to the heavy blow reported by the participant) was not associated with a change in TSSS.

Table 6.4. Cumulative burden for each athlete-session and each metric, calculated as the sum of all individual video-verified, 'good' mouthguards recordings associated with a head impact.

	Symptoms	Head impact events			Cumulative burden									
Athlete-session	Max change in TSSS	Video events[a]	All VV MG[b]	'Good' MG[c]	PLA (g)	PAV ($rad.s^{-1}$)	PAA ($rad.s^{-2}$)	$\Delta\omega_{max}$ ($rad.s^{-1}$)	GSI	HIC_{36}	RIC_{36}	BrIC	GAMBIT	HIP (W)
MG03														
Session 1	-1	481	98	44	842.3	508.1	88059.9	486.3	414.2	371.8	4.35E+07	10.6	12.2	31432.3
MG04														
Session 1*	13	380	63	24	299.2	309	33476.8	273.5	110.7	113.7	1.21E+07	5.8	4.5	12328.4
Session 2	-1	204	19	6	117.9	82.2	14574.5	76.5	78.4	71.5	9.86E+06	1.8	1.9	4792.9
Session 3	-4	319	33	10	120.1	125.7	13735.8	91	33.6	29.3	3.85E+06	2	1.9	3518.8
MG05														
Session 1	-1	479	94	48	837.6	714.5	108134.7	680.5	503.9	490.3	6.65E+07	14.4	14.2	44093.3
Session 2	0	290	52	33	552.8	454.4	60408.9	408.9	257.3	221.2	2.54E+07	8.8	8.3	20026.9
Session 3	0	255	31	16	211.6	205.6	28486.1	179	82.5	78.2	1.01E+07	3.7	3.7	8737.7
MG07														
Session 1	-1	335	59	21	378.1	336	51062.7	311.2	186	185.9	3.30E+07	6.6	6.6	16867.6
Session 2*	17	208	48	21	360	297	39198.4	251.9	148.9	127.3	1.46E+07	5.4	5.4	10534.6
Session 3*	5	276	84	37	637.3	498.3	73664.6	452.2	344	304.1	3.83E+07	9.8	9.9	23306.2
MG08														
Session 1*	13	231	51	28	433.9	377.3	49079.8	339.2	208.9	195.5	2.07E+07	7.2	6.6	18927.6
MG09														
Session 1	0	409	82	58	1159.1	693.2	84777.9	647.4	761.4	655.9	3.10E+07	12.8	13.1	55178.2
Session 2	0	524	84	49	744.4	537.6	60789.5	469	324.4	293.3	1.61E+07	9.8	9.1	26525.4
Session 3	0	91	14	6	78.9	69	8259.4	51.6	19.7	18.2	1.89E+06	1.1	1.1	1750.9
MG10														
Session 1	-1	486	112	62	1064.6	758.2	94350.3	761.5	622.1	535	3.84E+07	15.7	13.7	51717.4
Session 2	-2	605	35	19	272.6	232.4	25291.0	211.3	110.6	99.3	7.45E+06	4.6	3.6	10352.8
Mean ± SD	2.3 ± 6.3	348.3 ± 140	59.9 ± 29.7	30.1 ± 17.9	506.9 ± 341.2	387.4 ± 220.5	52084.4 ± 31088.4	355.7 ± 217.5	262.9 ± 218.4	236.9 ± 191.3	2.33E+07 ± 1.75E+07	7.5 ± 4.5	7.2 ± 4.4	21255.7 ± 16729.6

* indicates the sessions where the participant reported an increase in total symptom severity score (TSSS) equal or greater than 5 between pre-sparring and any time point post-sparring.
[a]Video events included all definite head impacts, prolonged contacts, probable head impacts, and clinches.
[b]All VV MG: included all the video-verified events recorded by the mouthguard and were associated with definite head impacts, prolonged contacts, probable head impacts, and clinches.
c'Good' MG: included only the video-verified mouthguard events that satisfied our quality criteria and were representative of true head motion.

Table 6.5. Maximum values measured for each athlete-session and each metric, out of all individual video-verified, 'good' mouthguards recordings associated with a head impact.

	Symptoms	Head impact events			Maximum value									
Athlete-session	Max change in TSSS	Video events[a]	All VV MG[b]	'Good' MG[c]	PLA (g)	PAV (rad.s^{-1})	PAA (rad.s^{-2})	$\Delta\omega_{max}$ (rad.s^{-1})	GSI	HIC_{36}	RIC_{36}	BrIC	GAMBIT	HIP (W)
MG03														
Session 1	-1	481	98	44	40.8	23.4	6760.1	23.4	43.4	35.5	7.80E+06	0.5	0.8	2379.8
MG04														
Session 1*	13	380	63	24	26.6	22.8	3251.5	23.1	23.3	22.0	2.53E+06	0.5	0.4	2307.6
Session 2	-1	204	19	6	41.4	21.8	5292.3	21.5	54.4	49.5	7.02E+06	0.5	0.7	2273.5
Session 3	-4	319	33	10	18.7	17.6	2375.0	15.3	9.7	8.2	9.35E+05	0.3	0.3	983.5
MG05														
Session 1	-1	479	94	48	38.3	29.7	5134.9	32.9	104.1	78.9	1.07E+07	0.6	0.7	5518.3
Session 2	0	290	52	33	38.0	21.6	4057.2	21.4	29.7	25.2	2.05E+06	0.4	0.5	1790.1
Session 3	0	255	31	16	24.1	18.7	3317.5	15.6	25.2	23.3	2.07E+06	0.4	0.4	2282.9
MG07														
Session 1	-1	335	59	21	32.1	30.2	5204.9	26.1	53.0	40.8	7.65E+06	0.6	0.7	3449.7
Session 2*	17	208	48	21	31.1	21.2	2778.3	20.3	18.0	16.0	1.63E+06	0.4	0.4	1173.3
Session 3*	5	276	84	37	37.7	30.8	5876.5	29.7	64.0	50.0	7.21E+06	0.7	0.7	3193.2
MG08														
Session 1*	13	231	51	28	28.4	20.5	3483.6	22.0	26.5	23.0	2.70E+06	0.5	0.5	2494.6
MG09														
Session 1	0	409	82	58	41.4	27.5	4107.7	25.8	45.7	39.9	2.26E+06	0.5	0.6	2798.0
Session 2	0	524	84	49	30.8	20.4	3386.0	19.4	27.0	22.7	2.75E+06	0.4	0.5	1561.8
Session 3	0	91	14	6	21.5	18.9	1632.0	11.2	8.6	7.5	5.37E+05	0.3	0.2	650.8
MG10														
Session 1	-1	486	112	62	43.0	22.7	3178.7	23.5	56.0	48.1	2.74E+06	0.5	0.4	3657.9
Session 2	-2	605	35	19	27.9	20.5	2575.0	20.3	24.3	19.9	1.28E+06	0.5	0.4	2010.4
Mean ± SD	2.3 ± 6.3	348.3 ± 140	59.9 ± 29.7	30.1 ± 17.9	32.6 ± 7.7	23 ± 4.2	3900.7 ± 1402.9	22 ± 5.4	38.3 ± 24.4	31.9 ± 18.6	3.87E+06 ± 3.10E+06	0.5 ± 0.1	0.5 ± 0.2	2407.8 ± 1184.3

* indicates the sessions where the participant reported an increase in total symptom severity score (TSSS) equal or greater than 5 between pre-sparring and any time point post-sparring.
[a]Video events included all definite head impacts, prolonged contacts, probable head impacts, and clinches.
[b]All VV MG: included all the video-verified events recorded by the mouthguard and were associated with definite head impacts, prolonged contacts, probable head impacts, and clinches.
[c]'Good' MG: included only the video-verified mouthguard events that satisfied our quality criteria and were representative of true head motion.

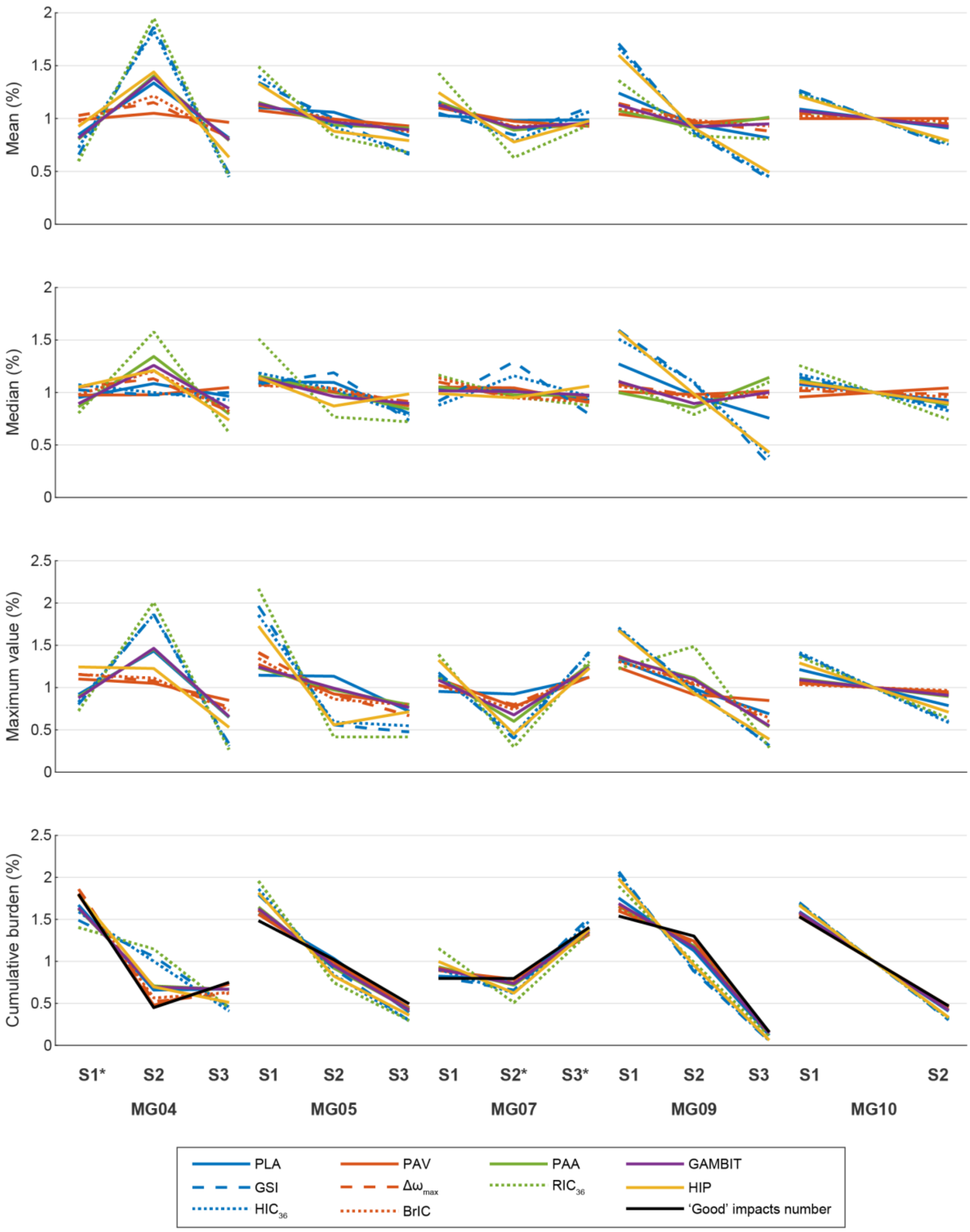
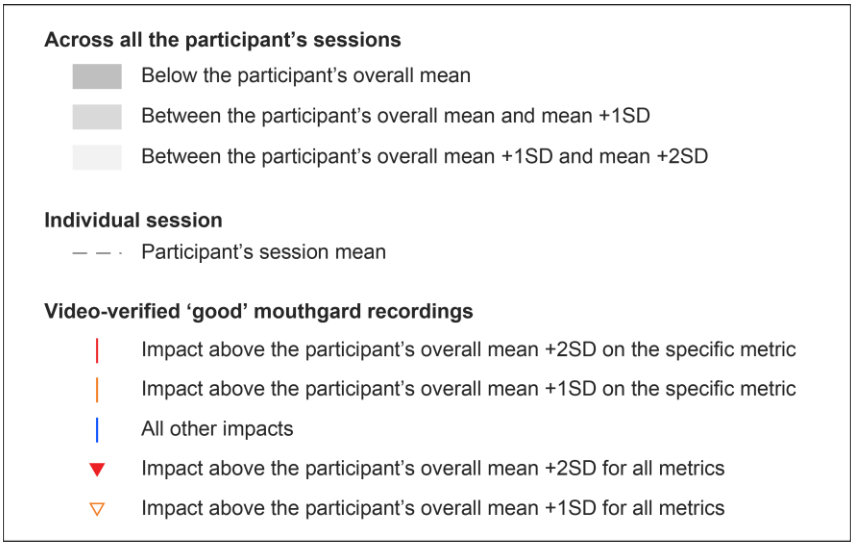


Figure 6.2. Variation in normalised mean, median, maximum value and cumulative burden across athlete-sessions for the ten calculated metrics. The number of 'good' mouthguard recordings is also represented on the cumulative burden plot (bottom row). Each value is expressed as a percentage of the participant's mean value calculated over the two or three athlete-sessions. Only the participants that were followed for more than one session are represented. The asterisks indicate the sessions where the participant reported an increase in total symptom severity score (TSSS) equal or greater than 5 between pre-sparring and any time point post-sparring.

From the control charts, 4 recordings (0.8% of all ‘good’ recordings) were of magnitude higher than two SDs of the participant’s mean on all ten metrics (Figure 6.3, Figure E.5). These four recordings included the heavy blow that was identified on video and met our quality criteria (MG04, Session 2) and impacts sustained by MG05 (Session 1), MG07 (Session 3), and MG09 (Session 1). Only one of those impacts occurred in a session after which the participant reported an increase in TSSS ≥ 5 (MG07, Session 3, increase of +5 between PRE and WOR). An additional 11 recordings (total of 15 or 3.1%) were of magnitude higher than one SD of the participant’s mean on all ten metrics, including only one recording sustained in an athlete-session after which an increase in TSSS was reported. The other heavy blows that were reported by the participants occurred to MG04 (Session 3, the recording did not satisfy our quality criteria) and MG08 (Session 1, could not be identified on video).

Figure 6.3. (next pages) Control charts: representation of the magnitude of all video-verified ‘good’ mouthguard recordings collected over three full athlete-sessions for participants MG04 and MG07. Each plot shows 70 minutes and covers the full sparring session. The upper border of the shaded areas represents the participants’ mean or the mean and one or two standard deviations (SD). Participant’s mean and SD values were calculated over all athlete-sessions available for the participant. Each vertical line represents one recording and is blue by default. The orange and red vertical lines represent the recordings for which the magnitude was above the participant’s mean +1SD (orange) or mean +2SD (red) values. The triangles highlight the recordings that were above the control limits for all 10 metrics. The asterisks by the session number (above the plots) indicate the sessions where the participant reported an increase in total symptom severity score ≥ 5 from pre- to post-sparring. Similar plots for the other participants are presented in Figure E.5.



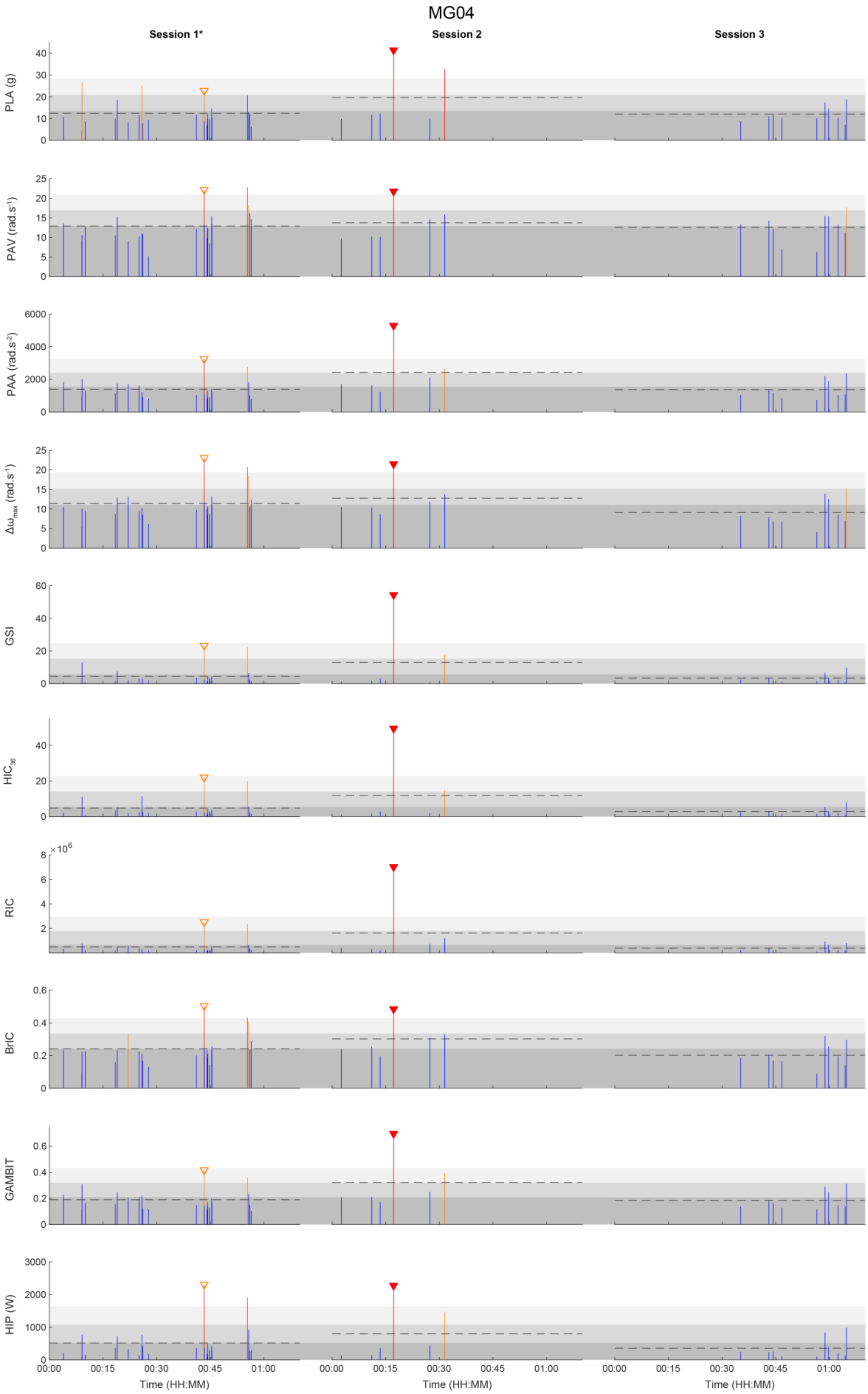


Figure 6.3. (continued)

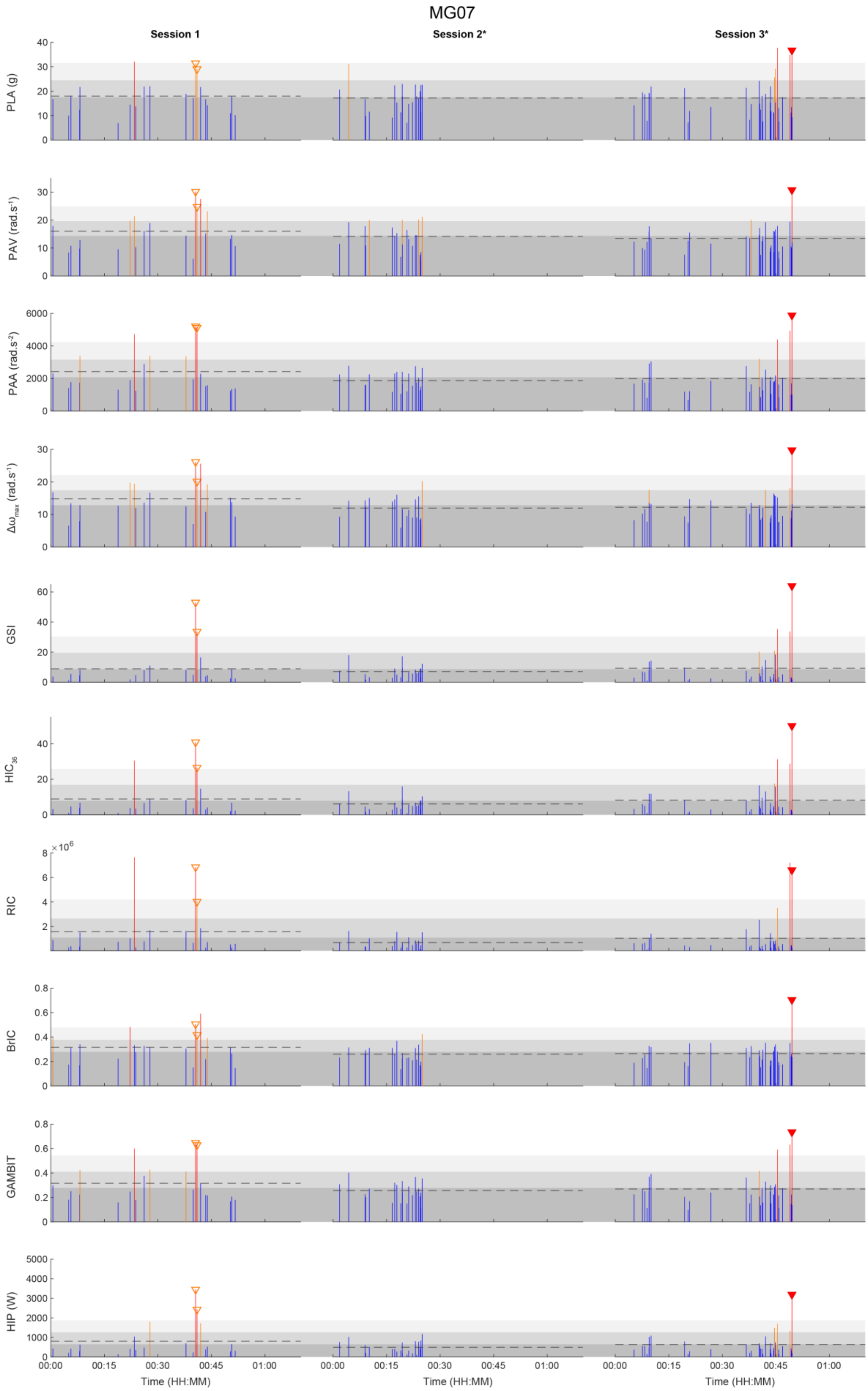


Figure 6.3. (continued)

6.6. Discussion

We monitored acute changes in concussion-related symptoms in seven boxers to evaluate four objectives: (1) to observe whether athletes experience acute changes in concussion-related symptoms after a session of boxing sparring; (2) to explore how these symptoms evolve over the subsequent 48 hours; (3) to qualitatively examine whether session-related exposure could be linked to acute changes in self-reported symptoms; and (4) to test the feasibility of using control charts to monitor exposure using individual-specific thresholds. Overall, we found that three out of seven participants self-reported changes in symptoms. In most of these cases, the symptom severity increased from pre- to post-sparring, then decreased over 48 hours, consistent with a typical injury recovery curve. The number of head impacts sustained during sparring varied across sessions, but the overall magnitude expressed by various kinematic metrics was in line with previously reported non-concussive impacts. Qualitatively, there did not seem to be consistency between session-specific cumulative burden and the magnitude of the change in symptom severity, but our small study size limits the certainty of this finding. Individual-specific control charts were used to identify head impacts presenting a magnitude larger than the average impact across a range of kinematic metrics. They correctly singled out a heavy blow that was reported by one of the participants but also highlighted several events sustained by participants who did not express any changes in symptomatology.

Acute symptoms changes resulting from sparring

When merging the results of the recruitment survey and the data collected over the study, there seems to be a risk group emerging from participants who reported in the survey experiencing heavy blows and symptoms, and reported acute changes in symptoms during the study, vs. participants who did not report anything. We note that these risk athletes came to the sport the most recently. It is possible that experienced boxers have habituated to the symptoms, or that a natural selection scenario occurs, where boxers that experience symptoms regularly do not stay in the sport. While the presence of different profiles needs to be further researched, the higher-risk athletes may need particular attention in terms of education and recovery management.[51]

From the results of the survey, experiencing symptoms more frequently may be associated with a higher number of estimated concussions, although a cause-consequence relationship cannot be inferred. Genetic factors,[146] co-morbidities such as chronic migraines or hyperactivity disorder,[379, 425] or a history of brain injury[256] may make some people more sensitive to experiencing symptoms, reflecting the concept of cognitive reserve.[183, 216] Furthermore, the symptoms are not specific to concussion and have been reported by patients with chronic pain, cervical spine and vestibular system dysfunction, depression, anxiety, and poor-quality sleep.[11, [51, 216, 252, 327] A much larger sample would be needed to assess the relationships between boxing experience, history of concussion and frequency of sparring-related symptoms. An online survey using the present study's questionnaire, independent of this thesis, is currently being disseminated to meet this end.

Over the whole cohort, there was no significant difference in the number of symptoms or the TSSS between pre- and post-sparring, consistent with the work of Kawata et al. on American football players.[190] However, when considering only the risk athletes, the proportion of athlete-sessions resulting in increases of TSSS ≥ 5 reached 50%. Three participants reported an increase in TSSS ranging from +5 to +17 (mean: 12 ± 5) between pre- and post-sparring over four athlete-sessions, resulting in a mean post-sparring TSSS of 19 ± 8 (using the highest value out of POST and WOR). These increases are similar to acute changes from baseline measured in concussed football players (mean change: 13 ± 9, range: 0-30),[149] and close to TSSS reported after concussion (median: 24, IQR: 11.5 – 45).[234, 352]

Whether the symptom score resulted directly from repetitive head impacts is not straightforward. Overall, it is common for athletes to report symptoms in the absence of injury.[11, 14] Such symptoms can result from physical activity and dehydration, heat-related illness, or anaemia.[324, 405] While unlikely to have been an issue for the participants in this study, the question of dehydration is particularly relevant in combat sports, where it can often be a result of weight management strategies.[89, 405] The participants also sometimes reported, prior to sparring, episodes of fatigue and poor sleep, illness, or neck/shoulder soreness; we eliminated this week-to-week within-subject variability[329] by calculating a session-specific pre/post change. Symptoms severity was also higher pre-sparring (Saturday mornings) than at 48H (Monday afternoons/evenings), which could be due to fatigue accumulated during the working/studying week. It may suggest that Saturday morning sparring sessions should be avoided in favour of another time where symptoms are less prevalent.

Evolution of symptoms over time

The symptoms experienced in this study appeared to be short-lived. Over the four instances where a change in TSSS ≥ 5 was recorded, the intensity was the highest immediately post-sparring (TSSS: 17 ± 12), then came down a few hours later (TSSS: 12 ± 2, 5 ± 1.5 hours after the end of the session), and was lower than pre-sparring at 48 hours (4 ± 3). This sequential course[256] is common following a concussion but occurs over a longer period (median TSSS of 26, 8, and 1 at 1, 5 and 11 days post-injury, respectively).[234] Because some of the symptoms reported immediately after sparring may be confounded with the effects of physical activity,[21, 324, 405] obtaining measures a few hours later may be more relevant to understanding the effect of repeated impacts to the head.

Implications for athlete safety

The time it takes for the symptoms to resolve may be considered a window of vulnerability,[33, 65] with the brain experiencing neurometabolic and neurovascular disturbances.[146, 342] The Association of Ringside Physicians states that "*Under no circumstances should a combat sports athlete compete or engage in sparring activity or competition if he or she is experiencing signs and symptoms of concussion.*"[291] As symptoms appear or get worse while sparring or fighting, the athlete's ability to protect themselves decreases via a decline in reaction time,[54] motor control,[106] or information retention,[378] consequently increasing the risks of getting hit and worsening the situation.[31] However, athletes will likely push through their symptoms while they are training, sparring, and especially fighting (preliminary results from our online survey), as well as return to training or competition shortly after having sustained a concussion, often hiding their symptoms and not seeking medical advice.[30, 51, 126]

Precautions are particularly important to take in the lead up to a fight, where the frequency and intensity of sparring tend to increase. Additionally, while the International Boxing Association (AIBA) rules limit the number of fights to one a day, it is common for amateur boxers to partake in multiple fights over a weekend or a couple of weeks (e.g., Olympic games). Suggestions to increase athlete safety include limiting the number and/or duration of rounds in a sparring session, increasing the duration of the breaks between rounds, or limiting the intensity of the strikes to the head. Until there exist objective tools that can assess an athlete's state in real-time, it is key that both athletes and trainers are educated on how to recognize the signs of concussion, and that they understand that removal from activity is the safest decision.

Association between session-related kinematics and acute changes in symptoms

The head acceleration events were of similar magnitude than non-concussive impacts reported in the literature for PLA (median: 15.5 g [IQR: 11.4 - 20.9]),[164, 220, 389, 416] PAA (1502.5 rad.s^{-2} [1070.3 - 2120.8]),[164, 220, 389, 416] $\Delta\omega_{max}$ (11 rad.s^{-1} [8.6 - 14.5]),[164, 416] BrIC (0.23 [0.18 - 0.3]),[164] and GAMBIT (0.21 [0.16 - 0.29]).[164] For the other metrics, the values recorded in the present study were an order of magnitude below previously reported non-injurious events: GSI (5.3 [2.5 - 10.1]),[164, 290, 378] HIC_{36} (4.9 [2.7 - 9.2]),[164, 290, 302, 378] RIC_{36} (4.2E+05 [2.1E+05 - 8.7E+05]),[164] and HIP (482.6 [275.4 - 903.7]).[164, 389] The highest magnitude events recorded for each athlete-session were also less than half the values reported as concussion tolerance threshold or concussive impacts for PLA (overall max: 43 g),[38, 423] GSI (104.1),[27, 229, 311] HIC_{36} (78.9),[27, 229, 311] and HIP (5.5 kW).[164, 294, 389]

However, several impacts were recorded with a magnitude close to concussive events for PAA (overall max: 6760.1 rad.s^{-2}),[38] $\Delta\omega_{max}$ (32.9 rad.s^{-1}),[164] RIC_{36} (1.1E+07),[164, 200] BrIC (0.7)[164] and GAMBIT (0.8).[164]

The small sample size of this feasibility study limits our ability to determine if there is an individual-specific relationship between any of the head impact measures (number of impacts, mean and median kinematics, cumulative burden or maximum value) and self-reported symptoms (number of symptoms, TSSS, feeling of normal, either post-sparring or the pre/post change). For participants MG04 and MG07, the session that resulted in the highest cumulative burdens also resulted in notable changes in TSSS (+13 and +5, respectively). On the other hand, the session leading to the largest change in TSSS (+17) had the lowest number of impacts and cumulative burden across the three athlete-sessions. This session may have been cut short after 6 rounds (the smallest number of rounds observed over the study) because of how the athlete was feeling (her anxiety/mood-related symptoms substantially increased between PRE and POST) and/or performing, but no information was given to the researcher from the athlete or the trainer.

To the best of our knowledge, there has not been any previous assessment of the relationship between cumulative burden and self-reported symptoms in the absence of a diagnosed concussion. One study[412] reported no association between the number of impacts sustained by Australian rules football players and baseline-to-post-game changes in symptomatology, although players were only assessed once and the analysis was conducted over the whole cohort. Another study[41] found no association between cumulative peak accelerations and symptoms for medically diagnosed concussed athletes but also had a unique data point for each participant. Research is needed to confirm or refute our preliminary findings, duplicating the key methodological aspects of our study, i.e., a cohort study design with pre/post assessments and an individual-specific investigation.

Our results also show that all metrics' cumulative burdens were correlated to each other and to the number of impacts, suggesting they were not more informative than the number of impacts in our analysis. The impacts' maximum values also showed some correlation across metrics. Cumulative burden as we calculated it assumes accumulation without accounting for recovery, while the maximum values account only for a few individual impacts, which were all sub-concussive in this study. It may be that none of these measures is representative of the brain's condition on its own. A previous study has shown promising preliminary results associating density-based metrics and concussion.[43] Further research is warranted to investigate metrics or analyses that account for the role of frequency and recovery on the occurrence of concussion-related symptoms.

Control charts to monitor exposure using participant-specific thresholds

We used a visual representation of the impacts over time inspired by control charts to identify outliers (i.e., events exceeding the upper-control limits across all ten metrics) for each participant. Our analysis showed that most outliers were not associated with an increase in symptoms. However, most of these events occurred to participants that never reported changes in symptoms, alluding that all the impacts sustained may have been below a theoretical individual tolerance threshold that was not reached during this study. Although anecdotal, it was promising that the only heavy blow available was singled out by the control charts analysis, even though the presence of this heavy blow was not associated with changes in symptoms.

Control charts are designed to evolve; with additional data, control limits may have to be adapted to better identify outliers (e.g., to three SDs over the mean). The control charts do not account for accumulation or a hypothetical decrease of injury tolerance as the number of impacts increases, but we could imagine using evolutive control limits declining along the session, rather than the fixed one chosen in this analysis. The control limits could also be adapted to the athlete's state pre-sparring to reflect the existence of symptoms. These methods are all articulated around finding an athlete-specific injury tolerance threshold and would aim at better identifying injured athletes or protecting individual athletes from injury. The potential of the control charts still needs to be investigated.

Limitations

The primary limitation of this feasibility study is the small sample size. We were bounded by equipment limitations and could only collect data from two participants for each session. By optimising the number and placement of cameras necessary to verify head impacts, and using appropriate and reliable sensors, we would be able to follow more participants each day.

Using a simple subjective symptom scale may not have provided an accurate representation of the participant's state, and it is typically used in parallel with more objective measures of neurocognitive testing involving memory, attention, and processing.[117, 256] The symptom scale was, however, the easiest and quickest way to assess the participant's state over several close time points, as each questionnaire took less than five minutes to complete and could be done remotely. We also hope that the questionnaires trained the participants in recognizing the signs that can be associated with a need for rest, should they experience them during or after training.

It appears that sleep problems, may be important in predisposing athletes for injury and recovery post-concussion.[11, 380] Our pre-sparring questionnaire did not include questions about the previous night's sleep quality and quantity, and future studies could include this aspect to evaluate its effect on symptoms changes.

We applied and evaluated ten metrics used in the prediction of (mild) traumatic brain injury. No single metric is recognised as the gold standard. These metrics have been developed from injurious data from mild to severe brain injuries, and their relevance in the study of sub-concussive head impacts has not been assessed. The metrics utilise different aspects of the kinematics signals (3DOF or resultant, linear or angular, acceleration or velocity), and different impacts were singled out by the control charts analysis depending on the metric used (i.e. the red and orange lines without the triangle). However, our findings also showed that all ten metrics were correlated when summed over full athlete-sessions, further correlated with impact count, and therefore provided redundant information. More research is needed into the clinical relevance of these metrics in the absence of a diagnosed concussion. We should also remember that 50% of the head impacts recorded by the mouthguards were discarded because their raw signals suggested they were not representative of the head's motion alone. Therefore, there may have been other impacts that would have reached the control limits, or the control limits may have changed and some of the outliers would not be identified as such.

6.7. Conclusion

This observational study aimed to explore the occurrence and longevity of concussion-related symptoms resulting from sparring and evaluate the potential use of individual-specific approaches for the monitoring of head impacts. Symptoms were observed in 25% of the sparring sessions overall, but in 50% for athletes in the identified risk group. The intensity of the symptoms immediately after the end of the sparring session was as high as post-concussion levels, but symptoms were short-lived and the overall intensity returned to equal or below pre-sparring levels within 48 hours. From our preliminary findings and with our small sample size, there was limited evidence that the accumulation of head impacts based on cumulative metrics, or the highest-magnitude impact sustained during a session, were associated with changes in self-reported symptoms. Symptoms may have resulted from the combination of several factors, namely a number of above-average-magnitude impacts, pre-existing dispositions (e.g., personal history and mental health, stress, poor-quality sleep, dehydration), and potentially a high density of head impacts. Our results support the concept of individual-specific analyses, as this approach allowed us to identify patterns emerging from a few athletes, that would not have been visible from cohort-averaged summaries. The existence of a risk group requires further research, as it may have implications for the identification of injurious impacts and the management of these athletes' training and recovery.

Chapter 7 - Discussion and conclusion

7.1. Key findings

The overall intention of this thesis was to explore the performance and limitations of some head impact sensors *in-vivo* so that we may better understand how these devices can be used reliably. This intention was motivated in part by the elusiveness of the hypothetical association between head impact kinematics and concussion. After summarizing 15 years of head impact research and identifying potential limitations, we assessed the performance and validity of an instrumented mouthguard in the laboratory under controlled impacts. Next, we compared an instrumented mouthguard, a skin patch and a patch attached to the headgear during boxing sparring. Finally, we evaluated the relationship between video-verified head impact kinematics and acute changes in self-reported symptoms over sparring sessions. The key findings of the thesis are:

- The number of false positive acceleration events being recorded by head impact sensors during boxing sparring was particularly low (< 1%), in contrast to reports from other sports (31-98%).
- The skin and headgear patches recorded twice as many video-verified acceleration events as the mouthguard during sparring despite being programmed to be triggered at the same linear acceleration threshold level of 10 g.
- The CSx sensors measured linear acceleration and angular velocity signals that adequately represented headform motion when tightly coupled to the headform in the laboratory (high agreement with the reference sensor in terms of signal shape), despite showing moderate accuracy.
- When used *in-vivo*, the CSx patches often (74-80%) recorded linear acceleration and angular velocity signals that did not satisfy our quality criteria (see Table 5.1, page 86) and were representative of skull/sensor decoupling. The Prevent Biometrics mouthguard, which was initially assumed to be well-coupled to the skull, showed a 50% proportion of such recordings.
- Impacts landing in the proximity of the sensor were associated with higher proportions of recordings being triggered and higher proportions of recordings that suggested skull/sensor decoupling.
- Low-quality recordings, identified by applying our quality criteria, measured generally higher peak linear and angular accelerations than high-quality recordings.

- There was little to no association in peak accelerations between the patches and the mouthguard among high-quality recordings.
- Overall, our boxers experienced acute changes in concussion-related symptoms from 25% of their sparring sessions, but from 50% of the sessions for athletes in a hypothetical risk group. These changes were seen mostly on the cognitive-fatigue symptoms cluster, were of intensity close to injury-related changes, and resolved within 48 hours.
- Accounting for our small sample size, there was limited evidence that the accumulation of head impacts based on cumulative metrics, or the highest-magnitude impact sustained during a session, were associated with changes in self-reported symptoms on an individual-specific basis.

Our findings highlight that to be able to use the kinematics from head impact sensors to study the effects of head impacts on an athlete's state and function, it is important that the sensors meet the following criteria:

1. High sensitivity: the sensor collects most of the head impacts that are potentially damaging.
2. Validity and accuracy: the resulting acceleration events are valid measurements of head motion, i.e., the kinematics represent head motion rather than sensor motion.
3. High specificity: if present, spurious recordings resulting from handling the sensor are properly excluded and low-quality recordings reflecting skull/sensor decoupling are properly identified.

While our methods did not allow us to conclude on criterion (1), the patches proved to be twice as sensitive (48-52%) to contacts to the head as the mouthguard (23%) in general, and even more sensitive to scoring punches (78-86% vs. 35%). However, according to criterion (2), both patches showed that only a small proportion (20-26%) of recordings reflected good coupling between the skull and the sensor. Furthermore, the anomalies observed in the patches' signals were mostly visible in the angular data, which have been suggested from animal and numerical studies to be the kinematic signals most strongly correlated with brain strains. It is therefore particularly important that angular kinematics be measured properly. The mouthguard was generally more rigidly attached to the skull, showing a 50-53% overall proportion of well-coupled recordings. Finally, our analyses found that the current systems and algorithms were not optimised to satisfy all of criterion (3). Although all sensors recorded few spurious events during sparring, the number of recordings misclassified as valid or invalid head impacts is concerning.

An overarching finding of this thesis pertains to the importance of skull/sensor coupling. Consistent with the literature, we observed physical decoupling in the case of the headgear and the skin patch and visualised the independent motion of the sensor on the raw kinematic traces for all three sensors. Tight coupling of the sensor to the skull is key to the sensors' ability to measure valid head impact exposure in terms of the number of impacts, magnitude, and direction of these impacts. Indeed, our results suggested that the number of head impacts may be overestimated as a sensor moves at a faster rate than the skull, triggering a recording while a better-fitted sensor may have measured sub-threshold accelerations. Particularly, the number of impacts may be inflated for impact location in close proximity to the sensor, creating a bias in the dataset. Furthermore, discarding low-quality signals potentially excludes impacts that participate in the alteration of an athlete's state, underestimating both the number of impacts sustained and the severity of the cumulative load. Although, if low-quality signals are not excluded from the analyses, then the overall severity of the exposure is overestimated because of the generally higher magnitude of these data compared with data from well-coupled sensors.

We can hypothesise that if a sensor were perfectly coupled to the skull, there wouldn't be any spurious events while the sensor is in place, and every event, whether on or off the field of play, would be a true head acceleration event. Additionally, there would not be any relative motion between the sensor and the skull, and 100% of the recorded kinematic signals would be representative of true head motion. The need for video verification, quality assessment or classification algorithm would be non-existent. Therefore, future research efforts in the head impact sensor field would be well spent on improving the mechanical coupling of the sensor to the skull. In reality, even with a tight fit, there will still be some variability and errors, which necessitate the continuing investment in classification algorithms.

The following conclusions can be drawn relative to the use of the three sensors assessed in our sparring study:

- Overall, for the analysis of acute exposure to head impacts, the mouthguard presented the best option out of the three sensors because of its better coupling to the skull. However, its capacity to record all the potentially injurious head impacts in boxing requires further investigation, and its moderate proportion of recordings with issues, together with the poor performance of the classification algorithm, suggest it should not be used as a "black box" for boxing head impacts.
- The patches did not provide enough data of acceptable quality to be useful for the analysis of head impact kinematics in boxing. Considering the consistency of our findings

with the recent literature regarding the coupling issues, caution is warranted for the use of patches and interpretation of patches' kinematic data in exposure and risk analyses.

- The headgear patch could be used as an impact counter in boxing because of its high sensitivity to head impacts and scoring punches, and because of its ease of use (no need to remove the headgear, no issues with sweating). As impacts to the back of the head only represented 2.5% of all events, the patch's location at the back of the head limits the effects of its sensitivity to impact location.
- Further work is needed to continue to refine these tools so that they more accurately record the number and magnitude of head impacts.

7.2. Limitations

Our findings are partly bound to the devices assessed in the thesis and their specifications:

- The duration of the recording, the turnover time between two recordings, the resolution of the timestamps, and the accuracy of the internal clock could have affected our capacity to match acceleration events to video events, and the validity of the comparison of acceleration events across sensors.
- It is possible that differences between the mouthguard and the patches, in the way data were processed and transformed to the head's centre of gravity, have participated in the differences in peak kinematics that we observed.
- Coupling to the skull may have been improved by using custom-fit mouthguards instead of boil-and-bite, or using stronger adhesive for the skin patches.
- Factors outside of our control forced us to change head impacts sensors between the laboratory and sparring studies. As a result, we assessed the performance of the CSx mouthguards in the laboratory but not *in-vivo*, and we deployed the Prevent Biometrics mouthguards and CSx patches in the sparring study without having appraised their performance in the laboratory. Laboratory validation of the Prevent Biometrics mouthguard and of the CSx patches with and with headgear was originally planned but had to be cancelled because of travel restrictions related to the COVID-19 pandemic. Nevertheless, Prevent Biometrics mouthguards have been previously validated and shown to have excellent accuracy in the laboratory, and the accelerometers and angular rate sensors in the CSx mouthguard are identical to those in the CSx patches, and therefore the laboratory experiment can be interpreted as a validation of the patches when tightly coupled to the headform.

- The Prevent Biometrics mouthguards, used exclusively to associate head impacts to symptoms of concussion in Chapter 6, have been validated by other research groups, however, we do not know how the specifics of boxing impacts (e.g., magnitude, direction, presence of headgear) influence the validity and accuracy of the measurements.

Some of our findings are directly associated with the sport of boxing and may be difficult to generalize to other sports.

- Impacts to the face, thus in proximity to the mouthguard, are common in boxing; similar work conducted in sports where impacts to the face are less frequent may show that the mouthguard performed better by recording fewer low-quality events overall. Still, our data showed that a proportion of 26-30% of recordings to the sides and back of the head did not meet our quality criteria.

- It is impossible to know from our data whether headgear interaction with the skin patch contributed to the number and magnitude of the acceleration events recorded by the skin patch. The skin patch may have shown better validity in the absence of headgear. Still, there was no association in peak kinematics between the two patches.

- We evaluated boxing headgear-based kinematic measurements but cannot assume that the results are generalisable to measurements from sensors embedded in hard-shell helmets, skull caps or headbands. However, given our concerning findings regarding the quality of the signals from the headgear patch, it would be beneficial to appraise the quality of the raw kinematics measured by sensors placed in these other types of headgear.

Finally, missing from our assessment of the sensors' performance is that we could not determine the risk that the sensors could miss important events. Our video coding methods were too inclusive to conduct such analysis, the number of heavy blows reported by the participants was too small, and we did not assess the visual severity of head impacts. Assessing the true sensitivity of head impact sensors is challenging as it is a function of the threshold used for the sensor, combined with the ability to determine the severity of an impact from video. Recording continuously would solve the issue but may be impractical with the current technology. Raters could be trained to visually classify head impacts below or above threshold from verified valid events, and consensus-based methods with several raters may be designed to increase the reliability of the classification.

7.3. Directions for future research

Methodological considerations for head impact kinematics

To satisfy criteria (2) and (3) described above, and because the current classification algorithms presented important limitations, several methodological steps are recommended. First, it is essential to remove all acceleration events that occur from the researcher and/or the athlete handling the sensor as these artefacts can affect both the number of events and the magnitude of peak kinematics summaries. Second, while false positives were rare during sparring and the positive predictive value of all sensors was above 96%, video-verifying each acceleration event is still the most rigorous approach, especially while we are in the research phase. Third, we recommend that the quality of each raw kinematic trace be visually assessed to identify the recordings that are evocative of skull/sensor decoupling.

Head impacts and signs of concussion

The value of head impact sensors for individual injury prevention remains in its infancy as it relies on a strong association between head kinematics and signs of concussion. Our findings related to the differences between sensors and the limited validity of the sensors for measuring *in-vivo* head motion may contribute to explaining why such association remains elusive. Additionally, we observed interesting differences in symptoms occurrence between participants and between sessions, highlighting several ideas:

- These differences would have remained hidden if we had not looked at each athlete-session specifically. Therefore, individual-specific methods, like control charts, are worth exploring as they might unravel whether tolerance thresholds to head impacts can be found on an individual basis in opposition to a global threshold.
- The assessment of symptoms changes between pre- and post-sparring may be more relevant to investigating the acute effects of head impacts than a comparison to baseline reports, that do not account for an athlete's state immediately before sports participation. Studies examining the concepts of acute cognitive reserve and pre-disposing factors would be worthwhile.
- The absence of clear associations between cumulative exposure or single high-magnitude impacts and changes in symptoms warrants the exploration of other dimensions of the exposure, such as the frequency or density of head impacts.
- As the focus of our explorations narrows (from population-wide exposure over a season of participation, to a single athlete-session), the validity and accuracy of the head impact

sensors to measure head motion become more important, and future research should be based on high-quality datasets.

7.4. Conclusions

The current thesis aimed to assess the performance of head impact sensors in the laboratory and *in-vivo*, to better understand how these devices can reliably be used in sports-related concussion research. The work has built upon and reinforced existing knowledge of the coupling issues between a head impact sensor and the skull, and these efforts have contributed a new and important understanding of the consequences of this critical research area. We offer recommendations on the use of head impact sensors, data processing methodology and analysis avenues, and hope that our work will help future research in the search of the association between head impact kinematics and concussion.

7.5. Open questions

As an indirect result of this thesis' work, two particular questions, related to past and future research, are intriguing:

- Given our findings of skull/sensor decoupling and the large range of concussive events magnitude in past research, how many of the high-range magnitudes were more representative of sensor decoupling than of true head motion?
- Finding an association between acute exposure to head impacts and signs of concussion would potentially allow preventing additional injury by removing an athlete after a potentially injurious head impact, and before compounding that injury with additional head impacts. In this association, head impact sensors play an intermediate role; instead, would it be possible to *directly* monitor an athlete's state (e.g., brain function or neurometabolic status) during sports participation?

References

[1]Aagaard J. and J. Du Bois. Telemetering impact data from the football field. *Electronics* 35: 46-47, 1962.
[2]Abrahams S., S. M. Fie, J. Patricios, M. Posthumus and A. V. September. Risk factors for sports concussion: an evidence-based systematic review. *British Journal of Sports Medicine* 48: 91-97, 2014.
[3]Allison M. A., Y. S. Kang, M. R. Maltese, J. H. t. Bolte and K. B. Arbogast. Measurement of Hybrid III head impact kinematics using an accelerometer and gyroscope system in ice hockey helmets. *Annals of Biomedical Engineering* 43: 1896-1906, 2015.
[4]Allison M. A., K. Yun Seok, J. B. Bolte Iv, M. R. Maltese and K. B. Arbogast. Validation of a helmet-based system to measure head impact biomechanics in ice hockey. *Medicine & Science in Sports & Exercise* 46: 115-123, 2014.
[5]Alois J., S. Bellamkonda, E. T. Campolettano, R. A. Gellner, A. Genemaras, J. G. Beckwith, . . . J. J. Crisco. Do American youth football players intentionally use their heads for high-magnitude impacts? *American Journal of Sports Medicine* 47: 3498-3504, 2019.
[6]Alosco M. L., J. Jarnagin, Y. Tripodis, B. Martin, C. Chaisson, C. M. Baugh, . . . R. A. Stern. Utility of providing a concussion definition in the assessment of concussion history in former NFL players. *Brain Injury* 1-8, 2017.
[7]Alosco M. L., Y. Tripodis, N. G. Fritts, A. Heslegrave, C. M. Baugh, S. Conneely, . . . R. A. Stern. Cerebrospinal fluid tau, Abeta, and sTREM2 in former National Football League players: Modeling the relationship between repetitive head impacts, microglial activation, and neurodegeneration. *Alzheimer's & Dementia* 14: 1159-1170, 2018.
[8]Alosco M. L., Y. Tripodis, J. Jarnagin, C. M. Baugh, B. Martin, C. E. Chaisson, . . . A. Jeromin. Repetitive head impact exposure and later-life plasma total tau in former National Football League players. *Alzheimer's & Dementia* 7: 33-40, 2017.
[9]Alsalaheen B., R. Bean, A. Almeida, J. Eckner and M. Lorincz. Characterization of cervical neuromuscular response to head-neck perturbation in active young adults. *Journal of Electromyography and Kinesiology* 39: 70-76, 2018.
[10]Asken B. M., Z. S. Brooke, T. C. Stevens, P. G. Silvestri, M. J. Graham, M. S. Jaffee and J. R. Clugston. Drill-specific head impacts in collegiate football practice: implications for reducing "Friendly Fire" exposure. *Annals of Biomedical Engineering* 47: 2094-2108, 2018.
[11]Asken B. M., A. R. Snyder, J. R. Clugston, L. S. Gaynor, M. J. Sullan and R. M. Bauer. Concussion-like symptom reporting in non-concussed collegiate athletes. *Archives of Clinical Neuropsychology* 32: 963-971, 2017.
[12]Badachhape A. A., R. J. Okamoto, C. L. Johnson and P. V. Bayly. Relationships between scalp, brain, and skull motion estimated using magnetic resonance elastography. *Journal of Biomechanics* 73: 40-49, 2018.
[13]Bahrami N., D. Sharma, S. Rosenthal, E. M. Davenport, J. E. Urban, B. Wagner, . . . J. A. Maldjian. Subconcussive head impact exposure and white matter tract changes over a single season of youth football. *Radiology* 281: 919-926, 2016.
[14]Balasundaram A. P., J. S. Sullivan, A. G. Schneiders and J. Athens. Symptom response following acute bouts of exercise in concussed and non-concussed individuals – A systematic narrative review. *Physical Therapy in Sport* 14: 253-258, 2013.
[15]Bari S., D. O. Svaldi, I. Jang, T. E. Shenk, V. N. Poole, T. Lee, . . . T. M. Talavage. Dependence on subconcussive impacts of brain metabolism in collision sport athletes: an MR spectroscopic study. *Brain Imaging and Behavior* 2018.
[16]Bartsch A., R. Dama, J. Alberts, S. Samorezov, E. Benzel, V. Miele, . . . B. Stemper. Measuring blunt force head impacts in athletes. *Military Medicine* 185: 190-196, 2020.
[17]Bartsch A., R. Dama, S. Samorezov, J. L. Alberts, E. Benzel, V. Miele, . . . B. Stemper. Laboratory and on-field results of athlete head impact monitoring. *Neurology* 91: S1-S1, 2018.
[18]Bartsch A., V. J. Miele, J. L. Alberts, E. Benzel, A. S. Shah, J. Humm, . . . M. McCrea. Estimates of high-risk single and cumulative head impact doses in American football. *Biomedical Sciences Instrumentation* 55: 224-229, 2019.
[19]Bartsch A., S. Samorezov, E. Benzel, V. Miele and D. Brett. Validation of an "Intelligent Mouthguard" single event head impact dosimeter. *Stapp Car Crash Journal* 58: 1-27, 2014.
[20]Bartsch A. J., D. S. Hedin, P. L. Gibson, V. J. Miele, E. C. Benzel, J. L. Alberts, . . . M. M. McCrea. Laboratory and on-field data collected by a head impact monitoring mouthguard. In: *Proceedings of the Annual International Conference of the IEEE Engineering in Medicine and Biology Society, EMBS*2019, p. 2068-2072.
[21]Basso J. C. and W. A. Suzuki. The effects of acute exercise on mood, cognition, neurophysiology, and neurochemical pathways: A review. *Brain Plasticity* 2: 127-152, 2017.
[22]Bauer R. and H. Fritz. Pathophysiology of traumatic injury in the developing brain: an introduction and short update. *Experimental and Toxicologic Pathology* 56: 65-73, 2004.
[23]Bazarian J. J., T. Zhu, J. Zhong, D. Janigro, E. Rozen, A. Roberts, . . . E. G. Blackman. Persistent, long-term cerebral white matter changes after sports-related repetitive head impacts. *PLoS ONE* 9: e94734, 2014.
[24]Beckwith J. G., J. J. Chu and R. M. Greenwald. Validation of a noninvasive system for measuring head acceleration for use during boxing competition. *Journal of Applied Biomechanics* 23: 238-244, 2007.
[25]Beckwith J. G., R. M. Greenwald and J. J. Chu. Measuring head kinematics in football: correlation between the head impact telemetry system and Hybrid III headform. *Annals of Biomedical Engineering* 40: 237-248, 2012.
[26]Beckwith J. G., R. M. Greenwald, J. J. Chu, J. J. Crisco, S. Rowson, S. M. Duma, . . . M. W. Collins. Head impact exposure sustained by football players on days of diagnosed concussion. *Medicine & Science in Sports & Exercise* 45: 737-746, 2013.

[27]Beckwith J. G., R. M. Greenwald, J. J. Chu, J. J. Crisco, S. Rowson, S. M. Duma, . . . M. W. Collins. Timing of concussion diagnosis is related to head impact exposure prior to injury. *Medicine & Science in Sports & Exercise* 45: 747-754, 2013.

[28]Beckwith J. G., W. Zhao, S. Ji, A. G. Ajamil, R. P. Bolander, J. J. Chu, . . . R. M. Greenwald. Estimated brain tissue response following impacts associated with and without diagnosed concussion. *Annals of Biomedical Engineering* 1-12, 2018.

[29]Beier G., E. Schuller, M. Schuck, C. Ewing, E. Becker and D. Thomas. Center of gravity and moments of inertia of human heads. 1980.

[30]Bennett L. L., J. J. Arias, P. J. Ford, C. Bernick and S. J. Banks. Concussion reporting and perceived knowledge of professional fighters. *The Physician and Sportsmedicine* 1-6, 2018.

[31]Bernick C., T. Hansen, W. Ng, V. Williams, M. Goodman, B. Nalepa, . . . T. Seifert. Concussion occurrence and recognition in professional boxing and MMA matches: toward a concussion protocol in combat sports. *The Physician and Sportsmedicine* 2020.

[32]Bland J. M. and D. G. Altman. Measuring agreement in method comparison studies. *Statistical methods in medical research* 8: 135-160, 1999.

[33]Blennow K., D. L. Brody, P. M. Kochanek, H. Levin, A. McKee, G. M. Ribbers, . . . H. Zetterberg. Traumatic brain injuries. *Nature Reviews Disease Primers* 2: 16084, 2016.

[34]Brainard L. L., J. G. Beckwith, J. J. Chu, J. J. Crisco, T. W. McAllister, A. C. Duhaime, . . . R. M. Greenwald. Gender differences in head impacts sustained by collegiate ice hockey players. *Medicine & Science in Sports & Exercise* 44: 297-304, 2012.

[35]Brauge D., C. Delpierre, P. Adam, J. C. Sol, P. Bernard and F.-E. Roux. Clinical and radiological cervical spine evaluation in retired professional rugby players. *Journal of Neurosurgery: Spine* 23: 551-557, 2015.

[36]Brault J. R., G. P. Siegmund and J. B. Wheeler. Cervical muscle response during whiplash: evidence of a lengthening muscle contraction. *Clinical Biomechanics* 15: 426-435, 2000.

[37]Breedlove E. L., M. Robinson, T. M. Talavage, K. E. Morigaki, U. Yoruk, K. O'Keefe, . . . E. A. Nauman. Biomechanical correlates of symptomatic and asymptomatic neurophysiological impairment in high school football. *Journal of Biomechanics* 45: 1265-1272, 2012.

[38]Brennan J. H., B. Mitra, A. Synnot, J. McKenzie, C. Willmott, A. S. McIntosh, . . . J. V. Rosenfeld. Accelerometers for the assessment of concussion in male athletes: a systematic review and meta-analysis. *Sports Medicine* 47: 469-478, 2016.

[39]Broglio S. P., J. T. Eckner, D. Martini, J. J. Sosnoff, J. S. Kutcher and C. Randolph. Cumulative head impact burden in high school football. *Journal of Neurotrauma* 28: 2069-2078, 2011.

[40]Broglio S. P., J. T. Eckner, H. L. Paulson and J. S. Kutcher. Cognitive decline and aging: the role of concussive and subconcussive impacts. *Exercise and Sport Sciences Reviews* 40: 138, 2012.

[41]Broglio S. P., J. T. Eckner, T. Surma and J. S. Kutcher. Post-concussion cognitive declines and symptomatology are not related to concussion biomechanics in high school football players. *Journal of Neurotrauma* 28: 2061-2068, 2011.

[42]Broglio S. P., K. M. Guskiewicz and J. Norwig. If you're not measuring, you're guessing: The advent of objective concussion assessments. *Journal of Athletic Training* 52: 160-166, 2017.

[43]Broglio S. P., A. P. Lapointe, K. L. O'Connor and M. McCrea. Head Impact Density: A model to explain the elusive concussion threshold. *Journal of Neurotrauma* 34: 2675-2683, 2017.

[44]Broglio S. P., S. N. Macciocchi and M. S. Ferrara. Sensitivity of the concussion assessment battery. *Neurosurgery* 60: 1050-1058, 2007.

[45]Broglio S. P., D. Martini, L. Kasper, J. T. Eckner and J. S. Kutcher. Estimation of head impact exposure in high school football: implications for regulating contact practices. *American Journal of Sports Medicine* 41: 2877-2884, 2013.

[46]Broglio S. P., B. Schnebel, J. J. Sosnoff, S. Shin, X. Fend, X. He and J. Zimmerman. Biomechanical properties of concussions in high school football. *Medicine & Science in Sports & Exercise* 42: 2064-2071, 2010.

[47]Broglio S. P., J. J. Sosnoff, S. Shin, X. He, C. Alcaraz and J. Zimmerman. Head impacts during high school football: a biomechanical assessment. *Journal of Athletic Training* 44: 342-349, 2009.

[48]Broglio S. P., R. Williams, A. Rettmann, B. Moore, J. T. Eckner and S. Meehan. No seasonal changes in cognitive functioning among high school football athletes: implementation of a novel electrophysiological measure and standard clinical measures. *Clinical Journal of Sport Medicine* 28: 130-138, 2017.

[49]Broglio S. P., R. M. Williams, K. L. O'Connor and J. Goldstick. Football players' head-impact exposure after limiting of full-contact practices. *Journal of Athletic Training* 51: 511-518, 2016.

[50]Brolinson P. G., S. Manoogian, D. McNeely, M. Goforth, R. Greenwald and S. Duma. Analysis of linear head accelerations from collegiate football impacts. *Current Sports Medicine Reports* 5: 23-38, 2006.

[51]Brown D. A., G. Grant, K. Evans, F. T. Leung and J. A. Hides. The association of concussion history and symptom presentation in combat sport athletes. *Physical Therapy in Sport* 48: 101-108, 2021.

[52]Buckley T. A., J. R. Oldham, D. J. Watson, N. G. Murray, B. A. Munkasy and K. M. Evans. Repetitive head impacts in football do not impair dynamic postural control. *Medicine & Science in Sports & Exercise* 51: 132-140, 2019.

[53]Buice J. M., A. O. Esquivel and C. J. Andrecovich. Laboratory validation of a wearable sensor for the measurement of head acceleration in men's and women's lacrosse. *Journal of Biomechanical Engineering* 140: 2018.

[54]Butler M. A., B. K. Doan, M. Hanna, G. A. Adam, A. Wile, B. Self, . . . E. Kryskow. An investigation of head accelerometry, cognitive function, and brain blood flow during intercollegiate boxing and its impact regarding head injury assessment in combat. Air Force Academy Colorado Springs, 2010.

[55]Caccese J. B., C. Best, L. C. Lamond, M. DiFabio, T. W. Kaminski, D. Watson, . . . T. A. Buckley. Effects of repetitive head impacts on a concussion assessment battery. *Medicine & Science in Sports & Exercise* 2019.

[56]Caccese J. B., L. C. Lamond, T. A. Buckley and T. W. Kaminski. Reducing purposeful headers from goal kicks and punts may reduce cumulative exposure to head acceleration. *Research in Sports Medicine* 24: 407-415, 2016.

[57]Camarillo D. B., P. B. Shull, J. Mattson, R. Shultz and D. Garza. An instrumented mouthguard for measuring linear and angular head impact kinematics in American football. *Annals of Biomedical Engineering* 41: 1939-1949, 2013.

[58]Campbell K. R., M. J. Warnica, I. C. Levine, J. S. Brooks, A. C. Laing, T. A. Burkhart and J. P. Dickey. Laboratory evaluation of the gForce Tracker™, a head impact kinematic measuring device for use in football helmets. *Annals of Biomedical Engineering* 44: 1246-1256, 2016.

[59]Campolettano E. T., G. Brolinson and S. Rowson. Postural control and head impact exposure in youth football players: Comparison of the Balance Error Scoring System and a force plate protocol. *Journal of Applied Biomechanics* 34: 127-133, 2017.

[60]Campolettano E. T., R. A. Gellner and S. Rowson. High-magnitude head impact exposure in youth football. *Journal of Neurosurgery. Pediatrics* 20: 604-612, 2017.

[61]Campolettano E. T., R. A. Gellner and S. Rowson. Relationship between impact velocity and resulting head accelerations during head impacts in youth football. *Conference proceedings International Research Council on the Biomechanics of Injury, IRCOBI* 2018-September: 326-333, 2018.

[62]Campolettano E. T., R. A. Gellner, E. P. Smith, S. Bellamkonda, C. T. Tierney, J. J. Crisco, . . . S. Rowson. Development of a concussion risk function for a youth population using head linear and rotational acceleration. *Annals of Biomedical Engineering* 2019.

[63]Campolettano E. T., S. Rowson and S. M. Duma. Drill-specific head impact exposure in youth football practice. *Journal of Neurosurgery. Pediatrics* 18: 536-541, 2016.

[64]Campolettano E. T., S. Rowson, S. M. Duma, B. Stemper, A. Shah, J. Harezlak, . . . M. McCrea. Factors affecting head impact exposure in college football practices: A multi-institutional study. *Annals of Biomedical Engineering* 2019.

[65]Cantu R. C. Second-impact syndrome. *Clinics in Sports Medicine* 17: 37-44, 1998.

[66]Caputo K. What I have learned from scoring 528 fights as a boxing judge. In: *The Guardian*2019.

[67]Carey L., P. Stanwell, D. P. Terry, A. S. McIntosh, S. V. Caswell, G. L. Iverson and A. J. Gardner. Verifying head impacts recorded by a wearable sensor using video footage in rugby league: a preliminary study. *Sports Medicine - Open* 5: 1-11, 2019.

[68]Caswell S. V., P. Kelshaw, A. E. Lincoln, L. Hepburn, R. Dunn and N. Cortes. Game-related impacts in high school boys' lacrosse. *Orthopaedic Journal of Sports Medicine* 7: 2019.

[69]Caswell S. V., A. E. Lincoln, H. Stone, P. Kelshaw, M. Putukian, L. Hepburn, . . . N. Cortes. Characterizing verified head impacts in high school girls' lacrosse. *American Journal of Sports Medicine* 45: 3374-3381, 2017.

[70]Cecchi N. J., D. C. Monroe, G. M. Fote, S. L. Small and J. W. Hicks. Head impacts sustained by male collegiate water polo athletes. *PLoS ONE* 14: 2019.

[71]Champagne A. A., N. S. Coverdale, M. Germuska, A. A. Bhogal and D. J. Cook. Changes in volumetric and metabolic parameters relate to differences in exposure to sub-concussive head impacts. *Journal of cerebral blood flow and metabolism* 1-15, 2019.

[72]Champagne A. A., N. S. Coverdale, J. Y. Nashed, J. Fernandez-Ruiz and D. J. Cook. Resting CMRO 2 fluctuations show persistent network hyper-connectivity following exposure to sub-concussive collisions. *NeuroImage: Clinical* 22: 2019.

[73]Champagne A. A., V. DiStefano, M. M. Boulanger, B. Magee, N. S. Coverdale, D. Gallucci, . . . D. J. Cook. Data-informed intervention improves football technique and reduces head impacts. *Medicine & Science in Sports & Exercise* 2019.

[74]Charek D. B., R. J. Elbin, A. Sufrinko, P. Schatz, N. R. D'Amico, M. W. Collins and A. P. Kontos. Preliminary evidence of a dose-response for continuing to play on recovery time after concussion. *Journal of Head Trauma Rehabilitation* 35: 85-91, 2020.

[75]Cheever K., K. Kawata, R. Tierney and A. Galgon. Cervical injury assessments for concussion evaluation: a review. *Journal of Athletic Training* 51: 1037-1044, 2016.

[76]Chrisman S. P., C. L. Mac Donald, S. Friedman, J. Andre, A. Rowhani-Rahbar, S. Drescher, . . . F. P. Rivara. Head impact exposure during a weekend youth soccer tournament. *Journal of Child Neurology* 31: 971-978, 2016.

[77]Chrisman S. P. D., B. E. Ebel, E. Stein, S. J. Lowry and F. P. Rivara. Head impact exposure in youth soccer and variation by age and sex. *Clinical Journal of Sport Medicine* 0: 1-8, 2017.

[78]Chun I. Y., X. Mao, E. L. Breedlove, L. J. Leverenz, E. A. Nauman and T. M. Talavage. DTI detection of longitudinal WM abnormalities due to accumulated head impacts. *Developmental Neuropsychology* 40: 92-97, 2015.

[79]Clark R. and A. G. Stanfill. A systematic review of barriers and facilitators for concussion reporting behavior among student athletes. *Journal of Trauma Nursing* 26: 297-311, 2019.

[80]Clay M. B., K. L. Glover and D. T. Lowe. Epidemiology of concussion in sport: a literature review. *Journal of Chiropractic Medicine* 12: 230-251, 2013.

[81]Cobb B. R., J. E. Urban, E. M. Davenport, S. Rowson, S. M. Duma, J. A. Maldjian, . . . J. D. Stitzel. Head impact exposure in youth football: elementary school ages 9-12 years and the effect of practice structure. *Annals of Biomedical Engineering* 41: 2463-2473, 2013.

[82]Collins C. L., E. N. Fletcher, S. K. Fields, L. Kluchurosky, M. K. Rohrkemper, R. D. Comstock and R. C. Cantu. Neck strength: a protective factor reducing risk for concussion in high school sports. *Journal of Primary Prevention* 35: 309-319, 2014.

[83]Cooper S.-M., M. Hughes, P. O'Donoghue and M. A. Nevill. A simple statistical method for assessing the reliability of data entered into sport performance analysis systems. *International Journal of Performance Analysis in Sport* 7: 87-109, 2007.
[84]Corna S., Y. Ito, M. von Brevern, A. M. Bronstein and M. A. Gresty. Reflex (unloading) and (defensive capitulation) responses in human neck muscle. *Journal of Physiology* 496 (Pt 2): 589-596, 1996.
[85]Corrigan J. D. and J. Bogner. Initial reliability and validity of the Ohio State University TBI identification method. *Journal of Head Trauma Rehabilitation* 22: 318-329, 2007.
[86]Cortes N., A. E. Lincoln, G. D. Myer, L. Hepburn, M. Higgins, M. Putukian and S. V. Caswell. Video analysis verification of head impact events measured by wearable sensors. *American Journal of Sports Medicine* 45: 2379-2387, 2017.
[87]Cournoyer J. and T. B. Hoshizaki. Head dynamic response and brain tissue deformation for boxing punches with and without loss of consciousness. *Clinical Biomechanics* 67: 96-101, 2019.
[88]Craton N. and O. Leslie. Time to re-think the Zurich Guidelines?: a critique on the consensus statement on concussion in sport: the 4th International Conference on Concussion in Sport, held in Zurich, November 2012. *Clinical Journal of Sport Medicine* 24: 93-95, 2014.
[89]Crighton B., G. L. Close and J. P. Morton. Alarming weight cutting behaviours in mixed martial arts: a cause for concern and a call for action. *British Journal of Sports Medicine* 50: 446-447, 2016.
[90]Crisco J. J., R. Fiore, J. G. Beckwith, J. J. Chu, P. G. Brolinson, S. Duma, . . . R. M. Greenwald. Frequency and location of head impact exposures in individual collegiate football players. *Journal of Athletic Training* 45: 549-459, 2010.
[91]Crisco J. J., B. J. Wilcox, J. G. Beckwith, J. J. Chu, A. C. Duhaime, S. Rowson, . . . R. M. Greenwald. Head impact exposure in collegiate football players. *Journal of Biomechanics* 44: 2673-2678, 2011.
[92]Crisco J. J., B. J. Wilcox, J. T. Machan, T. W. McAllister, A. C. Duhaime, S. M. Duma, . . . R. M. Greenwald. Magnitude of head impact exposures in individual collegiate football players. *Journal of Applied Biomechanics* 28: 174-183, 2012.
[93]Cummiskey B., D. Schiffmiller, T. M. Talavage, L. Leverenz, J. J. Meyer, D. Adams and E. A. Nauman. Reliability and accuracy of helmet-mounted and head-mounted devices used to measure head accelerations. *Proceedings of the Institution of Mechanical Engineers, Part P: Journal of Sports Engineering and Technology* 231: 144-153, 2017.
[94]Cummiskey B. R. Characterization and evaluation of head impact sensors and varsity football helmets. Purdue University, 2015.
[95]Curran-Sills G. and T. Abedin. Risk factors associated with injury and concussion in sanctioned amateur and professional mixed martial arts bouts in Calgary, Alberta. *BMJ Open Sport & Exercise Medicine* 4: 2018.
[96]Daneshvar D. H., C. J. Nowinski, A. C. McKee and R. C. Cantu. The epidemiology of sport-related concussion. *Clinics in Sports Medicine* 30: 1-17, 2011.
[97]Daniel R. W., S. Rowson and S. M. Duma. Head impact exposure in youth football. *Annals of Biomedical Engineering* 40: 976-981, 2012.
[98]Daniel R. W., S. Rowson and S. M. Duma. Head impact exposure in youth football: middle school ages 12-14 years. *Journal of Biomechanical Engineering* 136: 094501, 2014.
[99]Davenport E. M., K. Apkarian, C. T. Whitlow, J. E. Urban, J. H. Jensen, E. Szuch, . . . J. A. Maldjian. Abnormalities in diffusional kurtosis metrics related to head impact exposure in a season of high school varsity football. *Journal of Neurotrauma* 33: 2133-2146, 2016.
[100]Davenport E. M., C. T. Whitlow, J. E. Urban, M. A. Espeland, Y. Jung, D. A. Rosenbaum, . . . J. A. Maldjian. Abnormal white matter integrity related to head impact exposure in a season of high school varsity football. *Journal of Neurotrauma* 31: 1617-1624, 2014.
[101]Davis G. A., M. Makdissi, P. Bloomfield, P. Clifton, R. J. Echemendia, É. C. Falvey, . . . P. McCrory. International study of video review of concussion in professional sports. *British Journal of Sports Medicine* bjsports-2018-099727, 2018.
[102]Debison-Larabie C. Examining the relationship between cervical anthropometrics, head kinematics and cervical muscle responses to sudden head perturbations in competitive ice hockey players. In: *Health Sciences in Kinesiology*University of Ontario Institute of Technology, 2016, p. 116.
[103]Decq P., N. Gault, M. Blandeau, T. Kerdraon, M. Berkal, A. ElHelou, . . . J.-C. Peyrin. Long-term consequences of recurrent sports concussion. *Acta Neurochirurgica* 158: 289-300, 2016.
[104]Delaney J. S., V. Puni and F. Rouah. Mechanisms of injury for concussions in university football, ice hockey, and soccer: a pilot study. *Clinical Journal of Sports Medicine* 16: 162-165, 2006.
[105]Dennison C. R., E. M. Macri and P. A. Cripton. Mechanisms of cervical spine injury in rugby union: is it premature to abandon hyperflexion as the main mechanism underpinning injury? *British Journal of Sports Medicine* 46: 545-549, 2012.
[106]Di Virgilio T. G., M. Ietswaart, L. Wilson, D. I. Donaldson and A. M. Hunter. Understanding the consequences of repetitive subconcussive head impacts in sport: Brain changes and dampened motor control are seen after boxing practice. *Frontiers in Human Neuroscience* 13: 294, 2019.
[107]Diakogeorgiou E. and T. L. Miyashita. Effect of head impact exposures on changes in cognitive testing. *Orthopaedic Journal of Sports Medicine* 6: 2325967118761031, 2018.
[108]Dick R., M. S. Ferrara, J. Agel, R. Courson, S. W. Marshall, M. J. Hanley and F. Reifsteck. Descriptive epidemiology of collegiate men's football injuries: National Collegiate Athletic Association Injury Surveillance System, 1988-1989 through 2003-2004. *Journal of Athletic Training* 42: 221-233, 2007.
[109]Dick R. W. Is there a gender difference in concussion incidence and outcomes? *British Journal of Sports Medicine* 43: i46-i50, 2009.

[110]Dickson T. J., S. Trathen, G. Waddington, F. A. Terwiel and D. Baltis. A human factors approach to snowsport safety: Novel research on pediatric participants' behaviors and head injury risk. *Applied Ergonomics* 53 Pt A: 79-86, 2016.
[111]Dickson T. J., G. Waddington, S. Trathen, D. Baltis and R. Adams. Technology applications to enhance understanding of realtime snowsport head accelerations. *Procedia Engineering* 60: 220-225, 2013.
[112]Dompier T. P., Z. Y. Kerr, S. W. Marshall and et al. Incidence of concussion during practice and games in youth, high school, and collegiate american football players. *JAMA Pediatrics* 169: 659-665, 2015.
[113]Drew M. K., J. Cook and C. F. Finch. Sports-related workload and injury risk: simply knowing the risks will not prevent injuries: Narrative review. *British Journal of Sports Medicine* 50: 1306-1308, 2016.
[114]Duhaime A. C., J. G. Beckwith, A. C. Maerlender, T. W. McAllister, J. J. Crisco, S. M. Duma, . . . R. M. Greenwald. Spectrum of acute clinical characteristics of diagnosed concussions in college athletes wearing instrumented helmets: clinical article. *Journal of Neurosurgery* 117: 1092-1099, 2012.
[115]Duma S. M., S. J. Manoogian, W. R. Bussone, P. G. Brolinson, M. W. Goforth, J. J. Donnenwerth, . . . J. J. Crisco. Analysis of real-time head accelerations in collegiate football players. *Clinical Journal of Sport Medicine* 15: 3-8, 2005.
[116]Duma S. M. and S. Rowson. Every Newton Hertz: a macro to micro approach to investigating brain injury. In: *IEEE Engineering in Medicine and Biology Society*. Minneapolis, Minnesota, USA: 2009, p. 1123-1126.
[117]Eckner J. T. and J. S. Kutcher. Concussion symptom scales and sideline assessment tools: A critical literature update. *Current Sports Medicine Reports* 9: 8-15, 2010.
[118]Eckner J. T., K. L. O'Connor, S. P. Broglio and J. A. Ashton-Miller. Comparison of head impact exposure between male and female high school ice hockey athletes. *American Journal of Sports Medicine* 46: 2253–2262, 2018.
[119]Eckner J. T., Y. K. Oh, M. S. Joshi, J. K. Richardson and J. A. Ashton-Miller. Effect of neck muscle strength and anticipatory cervical muscle activation on the kinematic response of the head to impulsive loads. *American Journal of Sports Medicine* 42: 566-576, 2014.
[120]Eckner J. T., M. Sabin, J. S. Kutcher and S. P. Broglio. No evidence for a cumulative impact effect on concussion injury threshold. *Journal of Neurotrauma* 28: 2079-2090, 2011.
[121]Elkin B. S., J. M. Elliott and G. P. Siegmund. Whiplash injury or concussion? A possible biomechanical explanation for concussion symptoms in some individuals following a rear-end collision. *Journal of Orthopaedic and Sports Physical Therapy* 46: 874-885, 2016.
[122]Emery C. A., A. M. Black, A. Kolstad, G. Martinez, A. Nettel-Aguirre, L. Engebretsen, . . . K. Schneider. What strategies can be used to effectively reduce the risk of concussion in sport? *British Journal of Sports Medicine* 2017.
[123]Emery C. A., J. Kang, I. Shrier, C. Goulet, B. E. Hagel, B. W. Benson, . . . W. H. Meeuwisse. Risk of injury associated with body checking among youth ice hockey players. *JAMA* 303: 2265-2272, 2010.
[124]Fanton M., L. Wu and D. Camarillo. Comment on “Frequency and magnitude of game-related head impacts in male contact sports athletes: A systematic review and meta-analysis”. *Sports Medicine* 50: 841-842, 2020.
[125]FIFA. Big count: 265 million playing football. In: *FIFA Magazine*, edited by M. KunzFIFA, 2007.
[126]Follmer B., A. A. Varga and E. P. Zehr. Understanding concussion knowledge and behavior among mixed martial arts, boxing, kickboxing, and Muay Thai athletes and coaches. *The Physician and Sportsmedicine* 1-7, 2020.
[127]Ford J. M., K. R. Campbell, C. B. Ford, K. E. Boyd, D. A. Padua and J. P. Mihalik. Can functional movement assessment predict football head impact biomechanics? *Medicine & Science in Sports & Exercise* 50: 1233-1240, 2017.
[128]Foust D. R. Cervical range of motion and dynamic response and strength of cervical muscles. *Proceedings of the 17th Stapp Car Crash Conference* 1973.
[129]Fukuda T., S. Koike, S. Miyakawa, H. Fujiya and Y. Yamamoto. Impact on the head during collisions between university American football players-focusing on the number of head impacts and linear head acceleration. *The Journal of Physical Fitness and Sports Medicine* 6: 241-249, 2017.
[130]Fukushima M., K. Kaneoka, K. Ono, M. Sakane, S. Ujihashi and N. Ochiai. Neck injury mechanisms during direct face impact. *Spine* 31: 903-908, 2006.
[131]Fuller C. W., J. H. Brooks, R. J. Cancea, J. Hall and S. P. Kemp. Contact events in rugby union and their propensity to cause injury. *British Journal of Sports Medicine* 41: 862-867, 2007.
[132]Fuller C. W., A. Taylor, S. P. T. Kemp and M. Raftery. Rugby World Cup 2015: World Rugby injury surveillance study. *British Journal of Sports Medicine* 51: 51-57, 2016.
[133]Fuller G. W., S. P. Kemp and P. Decq. The International Rugby Board (IRB) pitch side concussion assessment trial: A pilot test accuracy study. *British Journal of Sports Medicine* bjsports-2014-093498, 2014.
[134]Funk J. R., S. M. Duma, S. J. Manoogian and S. Rowson. Biomechanical risk estimates for mild traumatic brain injury. *Association for the Advancement of Automotive Medicine Annual proceedings* 51: 343-361, 2007.
[135]Funk J. R., S. Rowson, R. W. Daniel and S. M. Duma. Validation of concussion risk curves for collegiate football players derived from HITS data. *Annals of Biomedical Engineering* 40: 79-89, 2012.
[136]Gabler L. F., S. H. Huddleston, N. Z. Dau, D. J. Lessley, K. B. Arbogast, X. Thompson, . . . J. R. Crandall. On-field performance of an instrumented mouthguard for detecting head impacts in American football. *Annals of Biomedical Engineering* 48: 2599-2612, 2020.
[137]Gadd C. W. Use of a weighted-impulse criterion for estimating injury hazard. SAE Technical Paper, 1966.
[138]Gallagher V., N. Kramer, K. Abbott, J. Alexander, H. Breiter, A. Herrold, . . . J. Reilly. The effects of sex differences and hormonal contraception on outcomes after collegiate sports-related concussion. *Journal of Neurotrauma* 35: 1242-1247, 2018.
[139]Garcia G.-G. P., C. Schumb, M. S. Lavieri, H. Koffijberg, T. McAllister, M. McCrea and S. P. Broglio. Developing insights for possible and probable acute concussions using cluster analysis. *Journal of Neurotrauma* 2021.

[140]Gavett B. E., R. A. Stern and A. C. McKee. Chronic traumatic encephalopathy: a potential late effect of sport-related concussive and subconcussive head trauma. *Clinics in Sports Medicine* 30: 179-188, 2011.
[141]Gehre C., H. Gades and P. Wernicke. Objective rating of signals using test and simulation responses. In: *Proceedings: International Technical Conference on the Enhanced Safety of Vehicles*National Highway Traffic Safety Administration, 2009.
[142]Gennarelli T., H. Segawa, U. Wald, Z. Czernicki, K. Marsh and C. Thompson. Physiological response to angular acceleration of the head. *Head injury: basic and clinical aspects* 1982: 129-140, 1982.
[143]Gennarelli T. A., L. E. Thibault, J. H. Adams, D. I. Graham, C. J. Thompson and R. P. Marcincin. Diffuse axonal injury and traumatic coma in the primate. *Annals of Neurology* 12: 564-574, 1982.
[144]Gilchrist I., K. Moglo, M. Storr and L. Pelland. Effects of head flexion posture on the multidirectional static force capacity of the neck. *Clinical Biomechanics* 37: 44-52, 2016.
[145]Gilchrist I., M. Storr, E. Chapman and L. Pelland. Neck muscle strength training in the risk management of concussion in contact sports: critical appraisal of application to practice. *Journal of Athletic Enhancement* 4: 19, 2015.
[146]Giza C. C. and D. A. Hovda. The new neurometabolic cascade of concussion. *Neurosurgery* 75: S24-S33, 2014.
[147]Greenwald R. M., J. T. Gwin, J. J. Chu and J. J. Crisco. Head impact severity measures for evaluating mild traumatic brain injury risk exposure. *Neurosurgery* 62: 789-798, 2008.
[148]Guskiewicz K. M. and J. P. Mihalik. Biomechanics of sport concussion: quest for the elusive injury threshold. *Exercise and Sport Sciences Reviews* 39: 4-11, 2011.
[149]Guskiewicz K. M., J. P. Mihalik, V. Shankar, S. W. Marshall, D. H. Crowell, S. M. Oliaro, . . . D. N. Hooker. Measurement of head impacts in collegiate football players: relationship between head impact biomechanics and acute clinical outcome after concussion. *Neurosurgery* 61: 1244-1253, 2007.
[150]Guskiewicz K. M. and T. C. Valovich McLeod. Pediatric sports-related concussion. *PM&R* 3: 353-364, 2011.
[151]Gwin J. T., J. J. Chu, S. G. Diamond, P. D. Halstead, J. J. Crisco and R. M. Greenwald. An investigation of the NOCSAE linear impactor test method based on in vivo measures of head impact acceleration in American Football. *Journal of Biomechanical Engineering* 132: 011006, 2010.
[152]Gwin J. T., J. J. Chu, T. A. McAllister and R. M. Greenwald. In situ measures of head impact acceleration in NCAA division I men's ice hockey: Implications for ASTM F1045 and other ice hockey helmet standards. *Journal of ASTM International* 6: 1-10, 2009.
[153]Gysland S. M., J. P. Mihalik, J. K. Register-Mihalik, S. C. Trulock, E. W. Shields and K. M. Guskiewicz. The relationship between subconcussive impacts and concussion history on clinical measures of neurologic function in collegiate football players. *Annals of Biomedical Engineering* 40: 14-22, 2012.
[154]Hanlon E. and C. Bir. Validation of a wireless head acceleration measurement system for use in soccer play. *Journal of Applied Biomechanics* 26: 424-431, 2010.
[155]Hanlon E. M. and C. A. Bir. Real-time head acceleration measurement in girls' youth soccer. *Medicine & Science in Sports & Exercise* 44: 1102-1108, 2012.
[156]Harper B., A. Aron, A. Siyufy and A. Mickle. Feasibility assessment of the Reebok CHECKLIGHT(TM) and King-Devick tests as screening tools in youth football. *Journal of Sports Medicine & Allied Health Sciences* 4: 1-7, 2018.
[157]Harpham J. A., J. P. Mihalik, A. C. Littleton, B. S. Frank and K. M. Guskiewicz. The effect of visual and sensory performance on head impact biomechanics in college football players. *Annals of Biomedical Engineering* 42: 1-10, 2014.
[158]Harriss A., A. M. Johnson, D. M. Walton and J. P. Dickey. Head impact magnitudes that occur from purposeful soccer heading depend on the game scenario and head impact location. *Musculoskeletal Science and Practice* 40: 53-57, 2019.
[159]Heath C. J. and J. L. Callahan. Self-reported concussion symptoms and training routines in mixed martial arts athletes. *Research in Sports Medicine* 21: 195-203, 2013.
[160]Hecimovich M., D. King, A. Dempsey, M. Gittins and M. Murphy. Youth Australian footballers experience similar impact forces to the head as junior- and senior-league players: A prospective study of kinematic measurements. *Journal of Sports Science And Medicine* 17: 547-556, 2018.
[161]Hecimovich M., D. King, A. Dempsey and M. Murphy. Head impact exposure in junior and adult Australian Football players. *Journal of Sports Medicine* Article ID 8376030: 2018.
[162]Hedin D. S., P. L. Gibson, A. J. Bartsch and S. Samorezov. Development of a head impact monitoring "Intelligent Mouthguard". In: *Conference proceedings: Annual International Conference of the IEEE Engineering in Medicine and Biology Society.*2016, p. 2007-2009.
[163]Heilbronner R. L., S. S. Bush, L. D. Ravdin, J. T. Barth, G. L. Iverson, R. M. Ruff, . . . D. K. Broshek. Neuropsychological consequences of boxing and recommendations to improve safety: A National Academy of Neuropsychology education paper. *Archives of Clinical Neuropsychology* 24: 11-19, 2009.
[164]Hernandez F., L. C. Wu, M. C. Yip, K. Laksari, A. R. Hoffman, J. R. Lopez, . . . D. B. Camarillo. Six degree-of-freedom measurements of human mild traumatic brain injury. *Annals of Biomedical Engineering* 43: 1918-1934, 2015.
[165]Higgins M., P. D. Halstead, L. Snyder-Mackler and D. Barlow. Measurement of impact acceleration: mouthpiece accelerometer versus helmet accelerometer. *Journal of Athletic Training* 42: 5-10, 2007.
[166]Hildenbrand K. J. and A. N. Vasavada. Collegiate and high school athlete neck strength in neutral and rotated postures. *Journal of Strength and Conditioning Research* 27: 3173-3182, 2013.
[167]Hirad A. A., J. J. Bazarian, K. Merchant-Borna, F. E. Garcea, S. Heilbronner, D. Paul, . . . B. Z. Mahon. A common neural signature of brain injury in concussion and subconcussion. *Science Advances* 5: eaau3460, 2019.
[168]Holbourn A. H. S. Mechanics of head injuries. *The Lancet* 242: 438-441, 1943.

[169]Huber B. R., M. L. Alosco, T. D. Stein and A. C. McKee. Potential long-term consequences of concussive and subconcussive injury. *Physical Medicine and Rehabilitation Clinics of North America* 27: 503-511, 2016.

[170]Huibregtse M. E., S. W. Zonner, K. Ejima, Z. W. Bevilacqua, S. D. Newman, J. T. Macy and K. Kawata. Association between muscle damage and head impacts in high school American football. *International Journal of Sports Medicine* 2019.

[171]Hume P. A., A. Theadom, G. N. Lewis, K. L. Quarrie, S. R. Brown, R. Hill and S. W. Marshall. A comparison of cognitive function in former rugby union players compared with former non-contact-sport players and the impact of concussion history. *Sports Medicine* 47: 1209-1220, 2016.

[172]Hunter L. E., C. A. Branch and M. L. Lipton. The neurobiological effects of repetitive head impacts in collision sports. *Neurobiology of Disease* 123: 122-126, 2018.

[173]Hurst H. T., S. Atkins and B. D. Dickinson. The magnitude of translational and rotational head accelerations experienced by riders during downhill mountain biking. *Journal of Science and Medicine in Sport* 1-6, 2018.

[174]Hurst H. T., L. Rylands, S. Atkins, K. Enright and S. J. Roberts. Profiling of translational and rotational head accelerations in youth BMX with and without neck brace. *Journal of Science and Medicine in Sport* 21: 263-267, 2017.

[175]Hynes L. M. and J. P. Dickey. Is there a relationship between whiplash-associated disorders and concussion in hockey? A preliminary study. *Brain Injury* 20: 179-188, 2006.

[176]International Boxing Association. AIBA referee and judge regulations. edited by Internation Boxing Association2018.

[177]Ito Y., S. Corna, M. von Brevern, A. Bronstein, J. Rothwell and M. Gresty. Neck muscle responses to abrupt free fall of the head: comparison of normal with labyrinthine-defective human subjects. *Journal of Physiology* 489: 911-916, 1995.

[178]Ito Y., S. Corna, M. vonBrevern, A. Bronstein and H. Gresty. The functional effectiveness of neck muscle reflexes for head-righting in response to sudden fall. *Experimental Brain Research* 117: 266-272, 1997.

[179]Iverson G. L., M. Gaetz, M. R. Lovell and M. W. Collins. Cumulative effects of concussion in amateur athletes. *Brain Injury* 18: 433-443, 2004.

[180]Jadischke R., D. C. Viano, N. Dau, A. I. King and J. McCarthy. On the accuracy of the Head Impact Telemetry (HIT) System used in football helmets. *Journal of Biomechanics* 46: 2310-2315, 2013.

[181]Jang I., I. Y. Chun, J. R. Brosch, S. Bari, Y. Zou, B. R. Cummiskey, . . . T. M. Talavage. Every hit matters: White matter diffusivity changes in high school football athletes are correlated with repetitive head acceleration event exposure. *NeuroImage: Clinical* 24: 1-16, 2019.

[182]Ji S., W. Zhao, Z. Li and T. W. McAllister. Head impact accelerations for brain strain-related responses in contact sports: a model-based investigation. *Biomechanics and Modeling in Mechanobiology* 13: 1121-1136, 2014.

[183]Jones R. N., J. Manly, M. M. Glymour, D. M. Rentz, A. L. Jefferson and Y. Stern. Conceptual and measurement challenges in research on cognitive reserve. *Journal of the International Neuropsychological Society* 17: 593-601, 2011.

[184]Joodaki H., A. Bailey, D. Lessley, J. Funk, C. Sherwood and J. Crandall. Relative motion between the helmet and the head in football impact test. *Journal of Biomechanical Engineering* 141: 2019.

[185]Jordan B. D., E. J. T. Matser, R. D. Zimmerman and T. Zazula. Sparring and cognitive function in professional boxers. *The Physician and Sportsmedicine* 24: 87-98, 1996.

[186]Jorgensen J. K., A. R. Thoreson, M. B. Stuart, A. Loyd, A. M. Smith, C. Twardowski, . . . M. J. Stuart. Interpreting oblique impact data from an accelerometer-instrumented ice hockey helmet. *Proceedings of the Institution of Mechanical Engineers, Part P: Journal of Sports Engineering and Technology* 231: 307-316, 2017.

[187]Joseph J. R., J. S. Swallow, K. Willsey, A. A. Almeida, M. T. Lorincz, R. K. Fraumann, . . . S. P. Broglio. Pupillary changes after clinically asymptomatic high-acceleration head impacts in high school football athletes. *Journal of Neurosurgery* 1-6, 2019.

[188]Joseph J. R., J. S. Swallow, K. Willsey, A. P. Lapointe, S. Khalatbari, F. K. Korley, . . . S. P. Broglio. Elevated markers of brain injury as a result of clinically asymptomatic high-acceleration head impacts in high-school football athletes. *Journal of Neurosurgery* 1-7, 2018.

[189]Kain D. J. It's just a concussion: The National Football League's denial of a casual link between multiple concussions and later-life cognitive decline. *Rutgers LJ* 40: 697, 2008.

[190]Kawata K., L. H. Rubin, J. H. Lee, T. Sim, M. Takahagi, V. Szwanki, . . . D. Langford. Association of football subconcussive head impacts with ocular near point of convergence. *JAMA Ophthalmology* 134: 763-769, 2016.

[191]Kawata K., L. H. Rubin, M. Takahagi, J. H. Lee, T. Sim, V. Szwanki, . . . D. Langford. Subconcussive impact-dependent increase in Plasma S100β levels in collegiate football players. *Journal of Neurotrauma* 34: 2254-2260, 2017.

[192]Kawata K., L. H. Rubin, L. Wesley, J. Lee, T. Sim, M. Takahagi, . . . D. Langford. Acute changes in plasma total Tau levels are independent of subconcussive head impacts in college football players. *Journal of Neurotrauma* 35: 260-266, 2017.

[193]Kelley M. E., J. M. Kane, M. A. Espeland, L. E. Miller, A. K. Powers, J. D. Stitzel and J. E. Urban. Head impact exposure measured in a single youth football team during practice drills. *Journal of Neurosurgery. Pediatrics* 20: 489-497, 2017.

[194]Kelley M. E., J. E. Urban, L. E. Miller, D. A. Jones, M. A. Espeland, E. M. Davenport, . . . J. D. Stitzel. Head impact exposure in youth football: Comparing age- and weight-based levels of play. *Journal of Neurotrauma* 34: 1939-1947, 2017.

[195]Kelshaw P., N. Cortes, A. Caswell and S. V. Caswell. Isometric cervical muscle strength does not affect head impact kinematics in high school boys' lacrosse. *International Journal of Athletic Therapy and Training* 23: 234-238, 2018.

[196]Kemp S. P., Z. Hudson, J. H. Brooks and C. W. Fuller. The epidemiology of head injuries in English professional rugby union. *Clinical Journal of Sports Medicine* 18: 227-234, 2008.
[197]Kieffer E. E. Sex-specific head impact exposure in rugby: Measurement considerations and relationships to clinical outcomes. Virginia Tech, 2021.
[198]Kieffer E. E., M. T. Begonia, A. M. Tyson and S. Rowson. A two-phased approach to quantifying head impact sensor accuracy: In-laboratory and on-field assessments. *Annals of Biomedical Engineering* 48: 2613-2625, 2020.
[199]Kieffer E. E., C. Vaillancourt, P. G. Brolinson and S. Rowson. Using in-mouth sensors to measure head kinematics in rugby. In: *International IRCOBI Conference on the Biomechanics of Injury*2020.
[200]Kimpara H. and M. Iwamoto. Mild traumatic brain injury predictors based on angular accelerations during impacts. *Annals of Biomedical Engineering* 40: 114-126, 2012.
[201]Kindschi K., M. Higgins, A. Hillman, G. Penczek and A. Lincoln. Video analysis of high-magnitude head impacts in men's collegiate lacrosse. *BMJ Open Sport & Exercise Medicine* 3: e000165, 2017.
[202]King A. I., K. H. Yang, L. Zhang, W. Hardy and D. C. Viano. Is head injury caused by linear or angular acceleration. In: *International IRCOBI Conference on the Biomechanics of Injury*2003, p. 1-12.
[203]King D., M. Hecimovich, T. Clark and C. Gissane. Measurement of the head impacts in a sub-elite Australian rules football team with an instrumented patch: an exploratory analysis. *International Journal of Sports Science & Coaching* 12: 359-370, 2017.
[204]King D., M. Hecimovich and T. Garrett. Accelerometric analysis of head impacts in amateur wrestling: an exploratory analysis. *International Journal of Wrestling Science* 6: 117-126, 2017.
[205]King D., P. Hume, C. Cummins, T. Clark, C. Gissane and M. Hecimovich. Head impact exposure comparison between male and female amateur rugby league participants measured with an instrumented patch. *Journal of sports medicine and therapy* 4: 024-037, 2019.
[206]King D., P. Hume, C. Gissane, M. Brughelli and T. Clark. The influence of head impact threshold for reporting data in contact and collision sports: systematic review and original data analysis. *Sports Medicine* 46: 151-169, 2016.
[207]King D., P. Hume, C. Gissane and T. Clark. Head impacts in a junior rugby league team measured with a wireless head impact sensor: an exploratory analysis. *Journal of Neurosurgery. Pediatrics* 19: 13-23, 2017.
[208]King D., P. Hume, C. Gissane, C. J. Cummins and Clark. Measurement of head impacts in a senior amateur rugby league team with an instrumented patch: exploratory analysis. *ARC Journal of Research in Sports Medicine* 2: 9-20, 2017.
[209]King D., P. A. Hume, M. Brughelli and C. Gissane. Instrumented mouthguard acceleration analyses for head impacts in amateur rugby union players over a season of matches. *American Journal of Sports Medicine* 43: 614-624, 2015.
[210]King D. A., P. A. Hume, C. Gissane and T. N. Clark. Similar head impact acceleration measured using instrumented ear patches in a junior rugby union team during matches in comparison with other sports. *Journal of Neurosurgery. Pediatrics* 18: 65-72, 2016.
[211]King D. A., P. A. Hume, C. Gissane, D. C. Kieser and T. N. Clark. Head impact exposure from match participation in women's rugby league over one season of domestic competition. *Journal of Science and Medicine in Sport* 21: 139-146, 2017.
[212]Kleiven S. Evaluation of head injury criteria using a finite element model validated against experiments on localized brain motion, intracerebral acceleration, and intracranial pressure. *International Journal of Crashworthiness* 11: 65-79, 2006.
[213]Kleiven S. Predictors for traumatic brain injuries evaluated through accident reconstructions. *Stapp Car Crash Journal* 51: 2007.
[214]Kleiven S. Predictors for traumatic brain injuries evaluated through accident reconstructions. *Stapp Car Crash Journal* 51: 81, 2007.
[215]Koh J. O., J. D. Cassidy and E. J. Watkinson. Incidence of concussion in contact sports: a systematic review of the evidence. *Brain Injury* 17: 901-917, 2003.
[216]Kontos A. P., A. Sufrinko, N. Sandel, K. Emami and M. W. Collins. Sport-related concussion clinical profiles: Clinical characteristics, targeted treatments, and preliminary evidence. *Current Sports Medicine Reports* 18: 82-92, 2019.
[217]Kumar S., R. Ferrari and Y. Narayan. Kinematic and electromyographic response to whiplash loading in low-velocity whiplash impacts—a review. *Clinical Biomechanics* 20: 343-356, 2005.
[218]Kumar S., Y. Narayan and T. Amell. Role of awareness in head-neck acceleration in low velocity rear-end impacts. *Accident Analysis & Prevention* 32: 233-241, 2000.
[219]Kuo C., J. Sganga, M. Fanton and D. B. Camarillo. Head impact kinematics estimation with network of inertial measurement units. *Journal of Biomechanical Engineering* 140: 2018.
[220]Kuo C., L. Wu, J. Loza, D. Senif, S. C. Anderson and D. B. Camarillo. Comparison of video-based and sensor-based head impact exposure. *PLoS ONE* 13: e0199238, 2018.
[221]Kuo C., L. Wu, W. Zhao, M. Fanton, S. Ji and D. B. Camarillo. Propagation of errors from skull kinematic measurements to finite element tissue responses. *Biomechanics and Modeling in Mechanobiology* 2017.
[222]Kuo C., L. C. Wu, B. T. Hammoor, J. F. Luck, H. C. Cutcliffe, R. C. Lynall, . . . D. B. Camarillo. Effect of the mandible on mouthguard measurements of head kinematics. *Journal of Biomechanics* 49: 1845-1853, 2016.
[223]Kuramochi R., T. Kimura, K. Nakazawa, M. Akai, S. Torii and S. Suzuki. Anticipatory modulation of neck muscle reflex responses induced by mechanical perturbations of the human forehead. *Neuroscience Letters* 366: 206-210, 2004.
[224]Lamond L. C., J. B. Caccese, T. A. Buckley, J. Glutting and T. W. Kaminski. Linear acceleration in direct head contact across impact type, player position, and playing scenario in collegiate women's soccer players. *Journal of Athletic Training* 53: 115-121, 2018.

[225]Le R. K., T. D. Saunders, K. M. Breedlove, D. A. Bradney, J. M. Lucas and T. G. Bowman. Differences in the mechanism of head impacts measured between men's and women's intercollegiate lacrosse athletes. *Orthopaedic Journal of Sports Medicine* 6: 1-7, 2018.

[226]Leeds D. D., C. D'Lauro and B. R. Johnson. Predictive power of head impact intensity measures for recognition memory performance. *Military Medicine* 184: 206-217, 2019.

[227]Lennon A. Measurement of head impact biomechanics: A comparison of the Head Impact Telemetry System and X2 Biosystems xPatch. University of North Carolina at Chapel Hill, 2015.

[228]Lewis L. M., R. Naunheim, J. Standeven, C. Lauryssen, C. Richter and B. Jeffords. Do football helmets reduce acceleration of impact in blunt head injuries? *Academic Emergency Medicine* 8: 604-609, 2001.

[229]Liao S., R. C. Lynall and J. P. Mihalik. The effect of head impact location on day of diagnosed concussion in college football. *Medicine & Science in Sports & Exercise* 48: 1239-1243, 2016.

[230]Limaye S. S., C. M. Mastrangelo and D. M. Zerr. A case study in monitoring hospital-associated infections with count control charts. *Quality Engineering* 20: 404-413, 2008.

[231]Lincoln A. E., S. V. Caswell, J. L. Almquist, R. E. Dunn, J. B. Norris and R. Y. Hinton. Trends in concussion incidence in high school sports: a prospective 11-year study. *American Journal of Sports Medicine* 39: 958-963, 2011.

[232]Liu Y., A. G. Domel, S. A. Yousefsani, J. Kondic, G. Grant, M. Zeineh and D. B. Camarillo. Validation and comparison of instrumented mouthguards for measuring head kinematics and assessing brain deformation in football impacts. *Annals of Biomedical Engineering* 48: 2580-2598, 2020.

[233]Llewellyn T., G. T. Burdette, A. B. Joyner and T. A. Buckley. Concussion reporting rates at the conclusion of an intercollegiate athletic career. *Clinical Journal of Sport Medicine* 24: 76-79, 2014.

[234]Lovell M. R., G. L. Iverson, M. W. Collins, K. Podell, K. M. Johnston, D. Pardini, . . . J. C. Maroon. Measurement of symptoms following sports-related concussion: Reliability and normative data for the Post-Concussion Scale. *Applied Neuropsychology* 13: 166-174, 2006.

[235]Lucas G. Q. A mechanical apparatus to quantify the reflex response of the human head/neck system. Washington State University, 2006.

[236]Lynall R. C., M. D. Clark, E. E. Grand, J. C. Stucker, A. C. Littleton, A. J. Aguilar, . . . J. P. Mihalik. Head impact biomechanics in women's college soccer. *Medicine & Science in Sports & Exercise* 48: 1772-1778, 2016.

[237]Lynall R. C., L. B. Lempke, R. S. Johnson, M. N. Anderson and J. D. Schmidt. A comparison of youth flag and tackle football head impact biomechanics. *Journal of Neurotrauma* 2019.

[238]Lystad R. P. and S. J. Strotmeyer. Concussion knowledge, attitudes and reporting intention among adult competitive Muay Thai kickboxing athletes: a cross-sectional study. *Injury Epidemiology* 5: 25, 2018.

[239]Mainwaring L., K. M. Ferdinand Pennock, S. Mylabathula and B. Z. Alavie. Subconcussive head impacts in sport: A systematic review of the evidence. *International Journal of Psychophysiology* 2018.

[240]Manley G., A. J. Gardner, K. J. Schneider, K. M. Guskiewicz, J. Bailes, R. C. Cantu, . . . G. L. Iverson. A systematic review of potential long-term effects of sport-related concussion. *British Journal of Sports Medicine* 51: 969-977, 2017.

[241]Mansell J., R. T. Tierney, M. R. Sitler, K. A. Swanik and D. Stearne. Resistance training and head-neck segment dynamic stabilization in male and female collegiate soccer players. *Journal of Athletic Training* 40: 310-319, 2005.

[242]Marchesseault E. R., D. Nguyen, L. Spahr, C. Beals, B. Razak and J. M. Rosene. Head impacts and cognitive performance in men's lacrosse. *The Physician and Sportsmedicine* 1-7, 2018.

[243]Margulies S. S. and L. E. Thibault. A proposed tolerance criterion for diffuse axonal injury in man. *Journal of Biomechanics* 25: 917-923, 1992.

[244]Martini D., J. Eckner, J. Kutcher and S. P. Broglio. Subconcussive head impact biomechanics: Comparing differing offensive schemes. *Medicine & Science in Sports & Exercise* 45: 755-761, 2013.

[245]Mathers C. H., J. P. Mihalik, D. M. Andrews, S. D. Watkins, P. Puzzuto and D. D. Kiecke. Using in-ear accelerometers to measure head acceleration in rough stock riders: A pilot study. *Athletic Training and Sports Health Care* 4: 158-164, 2012.

[246]Matheson L., V. Mooney, V. Caiozzo, G. Jarvis, J. Pottinger, C. DeBerry, . . . J. Antoni. Effect of instructions on isokinetic trunk strength testing variability, reliability, absolute value, and predictive validity. *Spine* 17: 914-921, 1992.

[247]McAllister T. and M. McCrea. Long-term cognitive and neuropsychiatric consequences of repetitive concussion and head-impact exposure. *Journal of Athletic Training* 52: 309-317, 2017.

[248]McAllister T. W., L. A. Flashman, A. Maerlender, R. M. Greenwald, J. G. Beckwith, T. D. Tosteson, . . . J. H. Turco. Cognitive effects of one season of head impacts in a cohort of collegiate contact sport athletes. *Neurology* 78: 1777-1784, 2012.

[249]McAllister T. W., J. C. Ford, L. A. Flashman, A. Maerlender, R. M. Greenwald, J. G. Beckwith, . . . S. Jain. Effect of head impacts on diffusivity measures in a cohort of collegiate contact sport athletes. *Neurology* 82: 63-69, 2014.

[250]McAllister T. W., J. C. Ford, S. Ji, J. G. Beckwith, L. A. Flashman, K. Paulsen and R. M. Greenwald. Maximum principal strain and strain rate associated with concussion diagnosis correlates with changes in corpus callosum white matter indices. *Annals of Biomedical Engineering* 40: 127-140, 2012.

[251]McCaffrey M. A., J. P. Mihalik, D. H. Crowell, E. W. Shields and K. M. Guskiewicz. Measurement of head impacts in collegiate football players: clinical measures of concussion after high- and low-magnitude impacts. *Neurosurgery* 61: 1236-1243, 2007.

[252]McClure D. J., S. L. Zuckerman, S. J. Kutscher, A. J. Gregory and G. S. Solomon. Baseline neurocognitive testing in sports-related concussions. *American Journal of Sports Medicine* 42: 472-478, 2014.

[253]McCrea M., K. Guskiewicz, C. Randolph, W. B. Barr, T. A. Hammeke, S. W. Marshall, . . . J. P. Kelly. Incidence, clinical course, and predictors of prolonged recovery time following sport-related concussion in high school and college athletes. *Journal of the International Neuropsychological Society* 19: 2013.
[254]McCrea M., T. Hammeke, G. Olsen, P. Leo and K. Guskiewicz. Unreported concussion in high school football players: implications for prevention. *Clinical Journal of Sport Medicine* 14: 13-17, 2004.
[255]McCrory P., N. Feddermann-Demont, J. Dvorak, J. D. Cassidy, A. McIntosh, P. E. Vos, . . . A. A. Tarnutzer. What is the definition of sports-related concussion: a systematic review. *British Journal of Sports Medicine* 51: 877-887, 2017.
[256]McCrory P., W. Meeuwisse, J. Dvorak, M. Aubry, J. Bailes, S. Broglio, . . . P. E. Vos. Consensus statement on concussion in sport—the 5th International Conference on Concussion in Sport held in Berlin, October 2016. *British Journal of Sports Medicine* 2017.
[257]McCuen E., D. Svaldi, K. Breedlove, N. Kraz, B. Cummiskey, E. L. Breedlove, . . . E. A. Nauman. Collegiate women's soccer players suffer greater cumulative head impacts than their high school counterparts. *Journal of Biomechanics* 48: 3720-3723, 2015.
[258]McIntosh A. S., C. Willmott, D. A. Patton, B. Mitra, J. H. Brennan, B. Dimech-Betancourt, . . . J. V. Rosenfeld. An assessment of the utility and functionality of wearable head impact sensors in Australian Football. *Journal of Science and Medicine in Sport* 22: 784-789, 2019.
[259]McMillan T. M., P. McSkimming, J. Wainman-Lefley, L. M. Maclean, J. Hay, A. McConnachie and W. Stewart. Long-term health outcomes after exposure to repeated concussion in elite level: rugby union players. *Journal of Neurology, Neurosurgery & Psychiatry* 2016.
[260]Meehan W. P., III, A. M. Taylor and M. Proctor. The pediatric athlete: younger athletes with sport-related concussion. *Clinics in Sports Medicine* 30: 133-144, 2011.
[261]Merchant-Borna K., P. Asselin, D. Narayan, B. Abar, C. M. Jones and J. J. Bazarian. Novel method of weighting cumulative helmet impacts improves correlation with brain white matter changes after one football season of sub-concussive head blows. *Annals of Biomedical Engineering* 44: 3679-3692, 2016.
[262]Michael D. B., D. R. Guyot and W. R. Darmody. Coincidence of head and cervical spine injury. *Journal of Neurotrauma* 6: 177-189, 1989.
[263]Mihalik J. P., D. R. Bell, S. W. Marshall and K. M. Guskiewicz. Measurement of head impacts in collegiate football players: an investigation of positional and event-type differences. *Neurosurgery* 61: 1229-1235, 2007.
[264]Mihalik J. P., J. T. Blackburn, R. M. Greenwald, R. C. Cantu, S. W. Marshall and K. M. Guskiewicz. Collision type and player anticipation affect head impact severity among youth ice hockey players. *Pediatrics* 125: e1394-e1401, 2010.
[265]Mihalik J. P., R. M. Greenwald, J. T. Blackburn, R. C. Cantu, S. W. Marshall and K. M. Guskiewicz. Effect of infraction type on head impact severity in youth ice hockey. *Medicine & Science in Sports & Exercise* 42: 1431-1438, 2010.
[266]Mihalik J. P., K. M. Guskiewicz, J. A. Jeffries, R. M. Greenwald and S. W. Marshall. Characteristics of head impacts sustained by youth ice hockey players. *Proceedings of the Institution of Mechanical Engineers, Part P: Journal of Sports Engineering and Technology* 222: 45-52, 2008.
[267]Mihalik J. P., K. M. Guskiewicz, S. W. Marshall, J. T. Blackburn, R. C. Cantu and R. M. Greenwald. Head impact biomechanics in youth hockey: Comparisons across playing position, event types, and impact locations. *Annals of Biomedical Engineering* 40: 141-149, 2012.
[268]Mihalik J. P., K. M. Guskiewicz, S. W. Marshall, R. M. Greenwald, J. T. Blackburn and R. C. Cantu. Does cervical muscle strength in youth ice hockey players affect head impact biomechanics? *Clinical Journal of Sport Medicine* 21: 416-421, 2011.
[269]Mihalik J. P., R. C. Lynall, E. B. Wasserman, K. M. Guskiewicz and S. W. Marshall. Evaluating the "Threshold Theory": can head impact indicators help? *Medicine & Science in Sports & Exercise* 49: 247-253, 2017.
[270]Mihalik J. P., A. Z. Sumrall, S. W. Yeargin, K. M. Guskiewicz, K. B. King, S. C. Trulock and E. W. Shields. Environmental and physiological factors affect football head impact biomechanics. *Medicine & Science in Sports & Exercise* 49: 2093-2101, 2017.
[271]Mihalik J. P., E. B. Wasserman, E. F. Teel and S. W. Marshall. Head impact biomechanics differ between girls and boys youth ice hockey players. *Annals of Biomedical Engineering* 2019.
[272]Miller L. E., C. Kuo, L. C. Wu, J. E. Urban, D. B. Camarillo and J. D. Stitzel. Validation of a custom instrumented retainer form factor for measuring linear and angular head impact kinematics. *Journal of Biomechanical Engineering* 140: 054501-054501-054506, 2018.
[273]Miller L. E., E. K. Pinkerton, K. C. Fabian, L. C. Wu, M. A. Espeland, L. C. Lamond, . . . J. E. Urban. Characterizing head impact exposure in youth female soccer with a custom-instrumented mouthpiece. *Research in Sports Medicine* 1-17, 2019.
[274]Miller L. E., J. E. Urban, M. E. Kelley, A. K. Powers, C. T. Whitlow, J. A. Maldjian, . . . J. D. Stitzel. Evaluation of brain response during head impact in youth athletes using an anatomically accurate finite element model. *Journal of Neurotrauma* 2018.
[275]Miyashita T. L., E. Diakogeorgiou and K. Marrie. Correlation of head impacts to change in Balance Error Scoring System scores in Division I men's lacrosse players. *Sports Health* 9: 318-323, 2017.
[276]Miyashita T. L., E. Diakogeorgiou and K. Marrie. The role of subconcussive impacts on sway velocities in Division I men's lacrosse players. *Sports Biomechanics* 1-9, 2018.
[277]Miyashita T. L. and P. A. Ullucci. Correlation of head impact exposures to vestibular ocular assessments. *Journal of Sport Rehabilitation* 1-19, 2019.

[278]Mohammed M. A., P. Worthington and W. H. Woodall. Plotting basic control charts: tutorial notes for healthcare practitioners. *Quality and Safety in Health Care* 17: 137-145, 2008.
[279]Moher D., L. Shamseer, M. Clarke, D. Ghersi, A. Liberati, M. Petticrew, . . . L. A. Stewart. Preferred reporting items for systematic review and meta-analysis protocols (PRISMA-P) 2015 statement. *Systematic Reviews* 4: 1-9, 2015.
[280]Montenigro P. H., M. L. Alosco, B. M. Martin, D. H. Daneshvar, J. Mez, C. E. Chaisson, . . . Y. Tripodis. Cumulative head impact exposure predicts later-life depression, apathy, executive dysfunction, and cognitive impairment in former high school and college football players. *Journal of Neurotrauma* 34: 328-340, 2017.
[281]Moon D. W., C. W. Beedle and C. R. Kovacic. Peak head acceleration of athletes during competition—football. *Medicine and Science in Sports* 3: 44-50, 1971.
[282]Muise D. P., S. J. MacKenzie and T. M. Sutherland. Frequency and magnitude of head accelerations in a Canadian interuniversity sport football teams training camp and season. *International Journal of Athletic Therapy & Training* 21: 36-41, 2016.
[283]Munce T. A., J. C. Dorman, P. A. Thompson, V. D. Valentine and M. F. Bergeron. Head impact exposure and neurologic function of youth football players. *Medicine & Science in Sports & Exercise* 47: 1567-1576, 2015.
[284]Myer G. D., K. Barber Foss, S. Thomas, R. Galloway, C. A. Dicesare, J. Dudley, . . . W. Yuan. Altered brain microstructure in association with repetitive subconcussive head impacts and the potential protective effect of jugular vein compression: A longitudinal study of female soccer athletes. *British Journal of Sports Medicine* 2018.
[285]Myer G. D., W. Yuan, K. D. Barber Foss, S. Thomas, D. Smith, J. Leach, . . . M. Altaye. Analysis of head impact exposure and brain microstructure response in a season-long application of a jugular vein compression collar: a prospective, neuroimaging investigation in American football. *British Journal of Sports Medicine* 50: 1276-1285, 2016.
[286]Nalepa B., A. Alexander, S. Schodrof, C. Bernick and J. Pardini. Fighting to keep a sport safe: toward a structured and sport-specific return to play protocol. *The Physician and Sportsmedicine* 45: 145-150, 2017.
[287]National Institute of Neurological Disorders and Stroke. NINDS Common Data Elements - Head kinematics estimates. https://commondataelements.ninds.nih.gov/: National Institute of Neurological Disorders and Stroke Common Data Elements.
[288]National Institute of Neurological Disorders and Stroke. NINDS Common Data Elements - Video device confirmation. https://commondataelements.ninds.nih.gov/: National Institute of Neurological Disorders and Stroke Common Data Elements.
[289]Nauman E. A., T. M. Talavage and P. S. Auerbach. Mitigating the consequences of subconcussive head injuries. *Annual Review of Biomedical Engineering* 22: 387-407, 2020.
[290]Naunheim R. S., J. Standeven, C. Richter and L. M. Lewis. Comparison of impact data in hockey, football, and soccer. *The Journal of Trauma: Injury, Infection, and Critical Care* 48: 938-941, 2000.
[291]Neidecker J., N. K. Sethi, R. Taylor, R. Monsell, D. Muzzi, B. Spizler, . . . C. Inalsingh. Concussion management in combat sports: consensus statement from the Association of Ringside Physicians. *British Journal of Sports Medicine* 53: 328-333, 2019.
[292]Nevins D., K. Hildenbrand, J. Kensrud, A. Vasavada and L. Smith. Field evaluation of a small form-factor head impact sensor for use in soccer. *Procedia Engineering* 147: 186-190, 2016.
[293]Newman J. A. A generalized acceleration model for brain injury threshold (GAMBIT). In: *International IRCOBI Conference on the Biomechanics of Injury*1986.
[294]Newman J. A. and N. Shewchenko. A proposed new biomechanical head injury assessment function - the Maximum Power Index. *Stapp Car Crash Journal* 44: 2000.
[295]Nijhuis L. B. O., J. H. Allum, G. F. Borm, F. Honegger, S. Overeem and B. R. Bloem. Directional sensitivity of “first trial” reactions in human balance control. *Journal of Neurophysiology* 101: 2802-2814, 2009.
[296]Noble J. M. and D. C. Hesdorffer. Sport-related concussions: a review of epidemiology, challenges in diagnosis, and potential risk factors. *Neuropsychology Review* 23: 273-284, 2013.
[297]O'Connor K. L., T. Peeters, S. Szymanski and S. P. Broglio. Individual impact magnitude vs. cumulative magnitude for estimating concussion odds. *Annals of Biomedical Engineering* 1-8, 2017.
[298]O'Connor K. L., S. Rowson, S. M. Duma and S. Broglio. Head-impact–measurement devices: a systematic review. *Journal of Athletic Training* 52: 206-227, 2017.
[299]O'Day K. M., E. M. Koehling, L. R. Vollavanh, D. Bradney, J. M. May, K. M. Breedlove, . . . T. G. Bowman. Comparison of head impact location during games and practices in Division III men's lacrosse players. *Clinical Biomechanics* 43: 23-27, 2017.
[300]O'Keeffe E., E. Kelly, Y. Liu, C. Giordano, E. Wallace, M. Hynes, . . . M. Campbell. Dynamic blood brain barrier regulation in mild head trauma. *Journal of Neurotrauma* 37: 347-356, 2019.
[301]O'Sullivan D., M. Roe and C. Blake. Analysis of head impacts during sub-elite hurling practice sessions. *Journal of Sports Sciences* 36: 1256-1261, 2017.
[302]O'Sullivan D. M. and G. P. Fife. Biomechanical head impact characteristics during sparring practice sessions in high school taekwondo athletes. *Journal of Neurosurgery. Pediatrics* 19: 662-667, 2017.
[303]O’Connell E. and M. Molloy. Concussion in rugby: knowledge and attitudes of players. *Irish Journal of Medical Science* 185: 521-528, 2016.
[304]Ocwieja K. E., J. P. Mihalik, S. W. Marshall, J. D. Schmidt, S. C. Trulock and K. M. Guskiewicz. The effect of play type and collision closing distance on head impact biomechanics. *Annals of Biomedical Engineering* 40: 90-96, 2012.
[305]Panjabi M. M., J. Cholewicki, K. Nibu, J. Grauer, L. B. Babat and J. Dvorak. Critical load of the human cervical spine: an in vitro experimental study. *Clinical Biomechanics* 13: 11-17, 1998.

[306]Patricios J., G. W. Fuller, R. Ellenbogen, S. Herring, J. S. Kutcher, M. Loosemore, . . . K. J. Schneider. What are the critical elements of sideline screening that can be used to establish the diagnosis of concussion? A systematic review. *British Journal of Sports Medicine* 51: 888-894, 2017.

[307]Patton D. A. A review of instrumented equipment to investigate head impacts in sport. *Applied Bionics and Biomechanics* Volume 2016, Article ID 7049743: 16 pages, 2016.

[308]Patton D. A., C. M. Huber, D. Jain, R. K. Myers, C. C. McDonald, S. S. Margulies, . . . K. B. Arbogast. Head impact sensor studies in sports: A systematic review of exposure confirmation methods. *Annals of Biomedical Engineering* 2020.

[309]Patton D. A., A. S. McIntosh and S. Kleiven. The biomechanical determinants of concussion: finite element simulations to investigate brain tissue deformations during sporting impacts to the unprotected head. *Journal of Applied Biomechanics* 29: 721-730, 2013.

[310]Pellman E. J., D. C. Viano, A. M. Tucker and I. R. Casson. Concussion in professional football: Location and direction of helmet impacts—Part 2. *Neurosurgery* 53: 1328-1341, 2003.

[311]Pellman E. J., D. C. Viano, A. M. Tucker, I. R. Casson and J. F. Waeckerle. Concussion in professional football: reconstruction of game impacts and injuries. *Neurosurgery* 53: 799-814, 2003.

[312]Penning L. Normal movements of the cervical spine. *American Journal of Roentgenology* 130: 317-326, 1978.

[313]Pfister T., K. Pfister, B. Hagel, W. A. Ghali and P. E. Ronksley. The incidence of concussion in youth sports: a systematic review and meta-analysis. *British Journal of Sports Medicine* 2015.

[314]Pincemaille Y., X. Trosseille, P. Mack, C. Tarriere, F. Breton and B. Renault. Some new data related to human tolerance obtained from volunteer boxers. *SAE transactions* 1752-1765, 1989.

[315]Pless I. B. and B. Hagel. Injury prevention: a glossary of terms. *Journal of Epidemiology & Community Health* 59: 182-185, 2005.

[316]Polinder S., M. C. Cnossen, R. G. L. Real, A. Covic, A. Gorbunova, D. C. Voormolen, . . . N. von Steinbuechel. A multidimensional approach to post-concussion symptoms in mild traumatic brain injury. *Frontiers in Neurology* 9: 2018.

[317]Popay J., H. Roberts, A. Sowden, M. Petticrew, L. Arai, M. Rodgers, . . . S. Duffy. Guidance on the conduct of narrative synthesis in systematic reviews. *A product from the ESRC methods programme Version* 1: b92, 2006.

[318]Portero R., J. Lecompte, P. Thoumie and P. Portero. Musculo-tendinous stiffness of the in vivo head-neck segment in response to quick-releases: A reproducibility study. *Isokinetics and Exercise Science* 19: 7-12, 2011.

[319]Portero R., F. Quaine, V. Cahouet, J. Lecompte, P. Thoumie and P. Portero. In vivo neck musculo-tendinous stiffness in response to quick-releases. In: *6th World Congress of Biomechanics (Wcb 2010)*2010, p. 593-596.

[320]Portero R., F. Quaine, V. Cahouet, P. Thoumie and P. Portero. Musculo-tendinous stiffness of head–neck segment in the sagittal plane: An optimization approach for modeling the cervical spine as a single-joint system. *Journal of Biomechanics* 46: 925-930, 2013.

[321]Post A. and T. B. Hoshizaki. Mechanisms of brain impact injuries and their prediction: A review. *Trauma* 14: 327-349, 2012.

[322]Press J. N. and S. Rowson. Quantifying head impact exposure in collegiate women's soccer. *Clinical Journal of Sport Medicine* 27: 104-110, 2017.

[323]Prien A., A. Grafe, R. Rössler, A. Junge and E. Verhagen. Epidemiology of head injuries focusing on concussions in team contact sports: a systematic review. *Sports Medicine* 48: 953-969, 2018.

[324]Putukian M. The acute symptoms of sport-related concussion: Diagnosis and on-field management. *Clinics in Sports Medicine* 30: 49-61, 2011.

[325]Quarrie K. L. and W. G. Hopkins. Tackle injuries in professional Rugby Union. *American Journal of Sports Medicine* 36: 1705-1716, 2008.

[326]Quarrie K. L. and I. R. Murphy. Towards an operational definition of sports concussion: identifying a limitation in the 2012 Zurich consensus statement and suggesting solutions. *British Journal of Sports Medicine* 48: 1589-1591, 2014.

[327]Rădoi A., M. A. Poca, D. Gándara, L. Castro, M. Cevallos, M. E. Pacios and J. Sahuquillo. The Sport Concussion Assessment Tool (SCAT2) for evaluating civilian mild traumatic brain injury. A pilot normative study. *PLoS ONE* 14: e0212541, 2019.

[328]Ralston J. D., J. Woodard, M. Cieslak, A. Asturias, W. Meiring and S. Grafton. A candidate neuromechanical biomarker and dosimeter for monitoring cumulative head impact trauma. In: *2017 IEEE EMBS International Conference on Biomedical & Health Informatics (BHI)*2017, p. 397-400.

[329]Rebchuk A. D., H. J. Brown, M. S. Koehle, J.-S. Blouin and G. P. Siegmund. Using variance to explore the diagnostic utility of baseline concussion testing. *Journal of Neurotrauma* 37: 1521-1527, 2020.

[330]Reed N., T. Taha, R. Greenwald and M. Keightley. Player and game characteristics and head impacts in female youth ice hockey players. *Journal of Athletic Training* 52: 771-775, 2017.

[331]Reed N., T. Taha, M. Keightley, C. Duggan, J. McAuliffe, J. Cubos, . . . W. Montelpare. Measurement of head impacts in youth ice hockey players. *International Journal of Sports Medicine* 31: 826-833, 2010.

[332]Reid S. E., G. Raviv and S. E. Reid, Jr. Neck muscle resistance to head impact. *Aviation Space and Environmental Medicine* 52: 78-84, 1981.

[333]Resch J. E., A. Rach, S. Walton and D. K. Broshek. Sport concussion and the female athlete. *Clinics in Sports Medicine* 36: 717-739, 2017.

[334]Reynolds B. B., J. Patrie, E. J. Henry, H. P. Goodkin, D. K. Broshek, M. Wintermark and T. J. Druzgal. Comparative analysis of head impact in contact and collision sports. *Journal of Neurotrauma* 34: 38-49, 2017.

[335]Reynolds B. B., J. Patrie, E. J. Henry, H. P. Goodkin, D. K. Broshek, M. Wintermark and T. J. Druzgal. Effects of sex and event type on head impact in collegiate soccer. *Orthopaedic Journal of Sports Medicine* 5: 2325967117701708, 2017.
[336]Reynolds B. B., J. Patrie, E. J. Henry, H. P. Goodkin, D. K. Broshek, M. Wintermark and T. J. Druzgal. Practice type effects on head impact in collegiate football. *Journal of Neurosurgery* 124: 501-510, 2016.
[337]Reynolds B. B., J. Patrie, E. J. Henry, H. P. Goodkin, D. K. Broshek, M. Wintermark and T. J. Druzgal. Quantifying head impacts in collegiate lacrosse. *American Journal of Sports Medicine* 44: 2947-2956, 2016.
[338]Rezasoltani A., J. Ylinen, A. Bakhtiary, M. Norozi and M. Montazeri. Cervical muscle strength measurement is dependent on the location of thoracic support. *British Journal of Sports Medicine* 42: 379-382, 2008.
[339]Rich A. M., T. M. Filben, L. E. Miller, B. T. Tomblin, A. R. Van Gorkom, M. A. Hurst, . . . J. D. Stitzel. Development, validation and pilot field deployment of a custom mouthpiece for head impact measurement. *Annals of Biomedical Engineering* 47: 2109-2121, 2019.
[340]Rihn J. A., D. T. Anderson, K. Lamb, P. F. Deluca, A. Bata, P. A. Marchetto, . . . A. R. Vaccaro. Cervical spine injuries in American football. *Sports Medicine* 39: 697-708, 2009.
[341]Robbins C. A., D. H. Daneshvar, J. D. Picano, B. E. Gavett, C. M. Baugh, D. O. Riley, . . . R. A. Stern. Self-reported concussion history: impact of providing a definition of concussion. *Open Access Journal of Sports Medicine* 5: 99-103, 2014.
[342]Romeu-Mejia R., C. C. Giza and J. T. Goldman. Concussion pathophysiology and injury biomechanics. *Current Reviews in Musculoskeletal Medicine* 12: 105-116, 2019.
[343]Rooks T. F., A. S. Dargie and V. C. Chancey. Machine learning classification of head impact sensor data. In: *ASME International Mechanical Engineering Congress and Exposition, Proceedings (IMECE)*2019.
[344]Rosene J. M., C. Merritt, N. R. Wirth and D. Nguyen. Incidence and force application of head impacts in men's lacrosse: A pilot study. *International Journal of Athletic Therapy and Training* 24: 213-216, 2019.
[345]Rowson S., J. G. Beckwith, J. J. Chu, D. S. Leonard, R. M. Greenwald and S. M. Duma. A six degree of freedom head acceleration measurement device for use in football. *Journal of Applied Biomechanics* 27: 8-14, 2011.
[346]Rowson S., G. Brolinson, M. Goforth, D. Dietter and S. Duma. Linear and angular head acceleration measurements in collegiate football. *Journal of Biomechanical Engineering* 131: 061016, 2009.
[347]Rowson S., E. T. Campolettano, S. M. Duma, B. Stemper, A. Shah, J. Harezlak, . . . M. McCrea. Concussion risk between individual football players: Survival analysis of recurrent events and non-events. *Annals of Biomedical Engineering* 48: 2626-2638, 2020.
[348]Rowson S., E. T. Campolettano, S. M. Duma, B. Stemper, A. Shah, J. Harezlak, . . . M. McCrea. Accounting for variance in concussion tolerance between individuals: Comparing head accelerations between concussed and physically matched control subjects. *Annals of Biomedical Engineering* 2019.
[349]Rowson S. and S. M. Duma. Brain injury prediction: Assessing the combined probability of concussion using linear and rotational head acceleration. *Annals of Biomedical Engineering* 41: 873-882, 2013.
[350]Rowson S. and S. M. Duma. Rotational acceleration and velocity associated with concussion in humans. In: *ASME 2011 Summer Bioengineering Conference*. Farmington, Pennsylvania, USA: 2011, p. 1039-1040.
[351]Rowson S., S. M. Duma, J. G. Beckwith, J. J. Chu, R. M. Greenwald, J. J. Crisco, . . . A. C. Maerlender. Rotational head kinematics in football impacts: an injury risk function for concussion. *Annals of Biomedical Engineering* 40: 1-13, 2012.
[352]Rowson S., S. M. Duma, B. D. Stemper, A. S. Shah, J. P. Mihalik, J. Harezlak, . . . M. McCrea. Correlation of concussion symptom profile with head impact biomechanics: a case for individual-specific injury tolerance. *Journal of Neurotrauma* 35: 681-690, 2017.
[353]Rubin L. H., R. Tierney, K. Kawata, L. Wesley, J. H. Lee, K. Blennow, . . . D. Langford. NFL blood levels are moderated by subconcussive impacts in a cohort of college football players. *Brain Injury* 33: 456-462, 2019.
[354]Salmon D. M., P. Handcock and B. Niven. Can neck strength be measured using a single maximal contraction in a simulated contact position? *Journal of Strength and Conditioning Research* 32: 2166-2173, 2017.
[355]Schmidt J. D., K. M. Guskiewicz, J. T. Blackburn, J. P. Mihalik, G. P. Siegmund and S. W. Marshall. The influence of cervical muscle characteristics on head impact biomechanics in football. *American Journal of Sports Medicine* 42: 2056-2066, 2014.
[356]Schmidt J. D., K. M. Guskiewicz, J. P. Mihalik, J. T. Blackburn, G. P. Siegmund and S. W. Marshall. Does visual performance influence head impact severity among high school football athletes? *Clinical Journal of Sport Medicine* 25: 494-501, 2015.
[357]Schmidt J. D., K. M. Guskiewicz, J. P. Mihalik, J. T. Blackburn, G. P. Siegmund and S. W. Marshall. Head impact magnitude in American high school football. *Pediatrics* 138: 2016.
[358]Schmidt J. D., T. T. Phan, R. W. Courson, F. I. Reifsteck, E. D. Merritt and C. N. Brown. The influence of heavier football helmet faceguards on head impact location and severity. *Clinical Journal of Sport Medicine* 28: 106-110, 2017.
[359]Schmidt J. D., A. F. Pierce, K. M. Guskiewicz, J. K. Register-Mihalik, D. N. Pamukoff and J. P. Mihalik. Safe-play knowledge, aggression, and head-impact biomechanics in adolescent ice hockey players. *Journal of Athletic Training* 51: 366-372, 2016.
[360]Schnebel B., J. T. Gwin, S. Anderson and R. Gatlin. In vivo study of head impacts in football: a comparison of National Collegiate Athletic Association Division I versus high school impacts. *Neurosurgery* 60: 490-496, 2007.
[361]Schneider D. K., R. Galloway, J. Bazarian, J. A. Diekfuss, J. Dudley, J. Leach, . . . G. D. Myer. Diffusion tensor imaging in athletes sustaining repetitive head impacts: A systematic review of prospective studies. *Journal of Neurotrauma* 36: 2831-2849, 2019.

[362]Schuh A., M. Canham-Chervak and B. H. Jones. Statistical process control charts for monitoring military injuries. *Injury Prevention* 23: 416-422, 2017.
[363]Schussler E., D. Stark, J. H. Bolte, Y. S. Kang and J. A. Onate. Comparison of a head mounted impact measurement device to the Hybrid III anthropomorphic testing device in a controlled laboratory setting. *International Journal of Sports Physical Therapy* 12: 592-600, 2017.
[364]Self B. P., C. Karins and T. Knox. Head accelerations during impact events. *Engineering of Sport* 2: 52-58, 2004.
[365]Shewchenko N., C. Withnall, M. Keown, R. Gittens and J. Dvorak. Heading in football. Part 1: Development of biomechanical methods to investigate head response. *British Journal of Sports Medicine* 39: i10-i25, 2005.
[366]Siegmund G. P., J.-S. Blouin and J. T. Inglis. Does startle explain the exaggerated first response to a transient perturbation? *Exercise and Sport Sciences Reviews* 36: 76-82, 2008.
[367]Siegmund G. P., S. J. Bonin, J. F. Luck and C. R. D. Bass. Validation of a skin-mounted sensor for measuring in-vivo head impacts. In: *International IRCOBI Conference on the Biomechanics of Injury*. Lyon, France: 2015, p. 182-183.
[368]Siegmund G. P., K. M. Guskiewicz, S. W. Marshall, A. L. Demarco and S. J. Bonin. A headform for testing helmet and mouthguard sensors that measure head impact severity in football players. *Annals of Biomedical Engineering* 42: 1834-1845, 2014.
[369]Siegmund G. P., K. M. Guskiewicz, S. W. Marshall, A. L. DeMarco and S. J. Bonin. Laboratory validation of two wearable sensor systems for measuring head impact severity in football players. *Annals of Biomedical Engineering* 44: 1257-1274, 2016.
[370]Siegmund G. P., D. J. Sanderson, B. S. Myers and J. T. Inglis. Rapid neck muscle adaptation alters the head kinematics of aware and unaware subjects undergoing multiple whiplash-like perturbations. *Journal of Biomechanics* 36: 473-482, 2003.
[371]Simoneau M., M. Denninger and T. C. Hain. Role of loading on head stability and effective neck stiffness and viscosity. *Journal of Biomechanics* 41: 2097-2103, 2008.
[372]Slobounov S. M., A. Walter, H. C. Breiter, D. C. Zhu, X. Bai, T. Bream, . . . T. M. Talavage. The effect of repetitive subconcussive collisions on brain integrity in collegiate football players over a single football season: A multi-modal neuroimaging study. *NeuroImage: Clinical* 14: 708-718, 2017.
[373]Stemper B. D., A. S. Shah, J. Harezlak, S. Rowson, S. Duma, J. P. Mihalik, . . . M. McCrea. Repetitive head impact exposure in college football following an NCAA rule change to eliminate two-a-day preseason practices: A study from the NCAA-DoD CARE Consortium. *Annals of Biomedical Engineering* 2019.
[374]Stemper B. D., A. S. Shah, J. Harezlak, S. Rowson, J. P. Mihalik, S. M. Duma, . . . J. P. DiFiori. Comparison of head impact exposure between concussed football athletes and matched controls: Evidence for a possible second mechanism of sport-related concussion. *Annals of Biomedical Engineering* 47: 2057-2072, 2019.
[375]Stemper B. D., A. S. Shah, A. Wild, J. R. Humm, F. A. Pintar, S. P. Broglio, . . . M. McCrea. Role of repetitive head impact exposure in the onset of concussion: Evidence of a possible second mechanism of concussion for contact sports. In: *Conference proceedings International Research Council on the Biomechanics of Injury, IRCOBI* 2018, p. 334-335.
[376]Stern R. A., D. O. Riley, D. H. Daneshvar, C. J. Nowinski, R. C. Cantu and A. C. McKee. Long-term consequences of repetitive brain trauma: Chronic traumatic encephalopathy. *PM&R* 3: S460-S467, 2011.
[377]Stiller J. W., S. S. Yu, L. A. Brenner, P. Langenberg, P. Scrofani, P. Pannella, . . . T. T. Postolache. Sparring and neurological function in professional boxers. *Frontiers in Public Health* 2: 2014.
[378]Stojsih S., M. Boitano, M. Wilhelm and C. Bir. A prospective study of punch biomechanics and cognitive function for amateur boxers. *British Journal of Sports Medicine* 44: 725-730, 2008.
[379]Sufrinko A., J. McAllister-Deitrick, R. Elbin, M. W. Collins and A. P. Kontos. Family history of migraine associated with posttraumatic migraine symptoms following sport-related concussion. *Journal of Head Trauma Rehabilitation* 33: 7-14, 2018.
[380]Sufrinko A., K. Pearce, R. J. Elbin, T. Covassin, E. Johnson, M. Collins and A. P. Kontos. The effect of preinjury sleep difficulties on neurocognitive impairment and symptoms after sport-related concussion. *American Journal of Sports Medicine* 43: 830-838, 2015.
[381]Svaldi D. O., C. Joshi, E. C. McCuen, J. P. Music, R. Hannemann, L. J. Leverenz, . . . T. M. Talavage. Accumulation of high magnitude acceleration events predicts cerebrovascular reactivity changes in female high school soccer athletes. *Brain Imaging and Behavior* 2018.
[382]Swain M. S., R. P. Lystad, H. Pollard and R. Bonello. Incidence and severity of neck injury in Rugby Union: A systematic review. *Journal of Science and Medicine in Sport* 14: 383-389, 2011.
[383]Swartz E. E., S. P. Broglio, S. B. Cook, R. C. Cantu, M. S. Ferrara, K. M. Guskiewicz and J. L. Myers. Early results of a helmetless-tackling intervention to decrease head impacts in football players. *Journal of Athletic Training* 50: 1219-1222, 2015.
[384]Swartz E. E., J. L. Myers, S. B. Cook, K. M. Guskiewicz, M. S. Ferrara, R. C. Cantu, . . . S. P. Broglio. A helmetless-tackling intervention in American football for decreasing head impact exposure: A randomized controlled trial. *Journal of Science and Medicine in Sport* 22: 1102-1107, 2019.
[385]Takhounts E. G., M. J. Craig, K. Moorhouse, J. McFadden and V. Hasija. Development of Brain Injury Criteria (BrIC). *Stapp Car Crash Journal* 57: 243-266, 2013.
[386]Talavage T. M., E. A. Nauman, E. L. Breedlove, U. Yoruk, A. E. Dye, K. E. Morigaki, . . . L. J. Leverenz. Functionally-detected cognitive impairment in high school football players without clinically-diagnosed concussion. *Journal of Neurotrauma* 31: 327-338, 2014.
[387]Thomson E., K. Lamb and C. Nicholas. The development of a reliable amateur boxing performance analysis template. *Journal of Sports Sciences* 31: 516-528, 2013.

[388]Tiernan S., G. Byrne and D. M. O'Sullivan. Evaluation of skin-mounted sensor for head impact measurement. *Proceedings of the Institution of Mechanical Engineers, Part H: Journal of Engineering in Medicine* 233: 735-744, 2019.
[389]Tiernan S., A. Meagher, D. O'Sullivan, E. O'Keeffe, E. Kelly, E. Wallace, . . . A. G. Domel. Concussion and the severity of head impacts in mixed martial arts. In: *Proceedings of the Institution of Mechanical Engineers*2020.
[390]Tierney G. J., J. Lawler, K. Denvir, K. McQuilkin and C. K. Simms. Risks associated with significant head impact events in elite rugby union. *Brain Injury* 1-12, 2016.
[391]Tierney R. T., M. R. Sitler, C. B. Swanik, K. A. Swanik, M. Higgins and J. Torg. Gender differences in head-neck segment dynamic stabilization during head acceleration. *Medicine & Science in Sports & Exercise* 37: 272-279, 2005.
[392]Triantafillou K. M., W. Lauerman and S. B. Kalantar. Degenerative disease of the cervical spine and its relationship to athletes. *Clinics in Sports Medicine* 31: 509-520, 2012.
[393]Trotta A., A. Ní Annaidh, R. O. Burek, B. Pelgrims and J. Ivens. Evaluation of the head-helmet sliding properties in an impact test. *Journal of Biomechanics* 75: 28-34, 2018.
[394]Trotta A., D. Zouzias, G. De Bruyne and A. Ní Annaidh. The importance of the scalp in head impact kinematics. *Annals of Biomedical Engineering* 46: 831-840, 2018.
[395]Tucker R., M. Raftery, S. Kemp, J. Brown, G. Fuller, B. Hester, . . . K. Quarrie. Risk factors for head injury events in professional rugby union: a video analysis of 464 head injury events to inform proposed injury prevention strategies. *British Journal of Sports Medicine* 51: 1152-1157, 2017.
[396]Tyson A. M., S. M. Duma and S. Rowson. Laboratory evaluation of low-cost wearable sensors for measuring head impacts in sports. *Journal of Applied Biomechanics* 1-24, 2018.
[397]Urban J. E., E. M. Davenport, A. J. Golman, J. A. Maldjian, C. T. Whitlow, A. K. Powers and J. D. Stitzel. Head impact exposure in youth football: high school ages 14 to 18 years and cumulative impact analysis. *Annals of Biomedical Engineering* 41: 2474-2487, 2013.
[398]Urban J. E., W. C. Flood, B. J. Zimmerman, M. E. Kelley, M. A. Espeland, L. McNamara, . . . J. D. Stitzel. Evaluation of head impact exposure measured from youth football game plays. *Journal of Neurosurgery. Pediatrics* 24: 190-199, 2019.
[399]Urban J. E., M. E. Kelley, M. A. Espeland, E. M. Davenport, C. T. Whitlow, A. K. Powers, . . . J. D. Stitzel. In-season variations in head impact exposure among youth football players. *Journal of Neurotrauma* 2018.
[400]Vandenbroucke J. P., E. Von Elm, D. G. Altman, P. C. Gøtzsche, C. D. Mulrow, S. J. Pocock, . . . S. Initiative. Strengthening the Reporting of Observational Studies in Epidemiology (STROBE): explanation and elaboration. *PLoS medicine* 4: e297, 2007.
[401]Vasavada A., D. Trask, A. Knottnerus and D. Lin. Effects of head position and impact direction on neck muscle response to perturbations. In: *American Society of Biomechanics Proceedings* 2009.
[402]Versace J. A review of the Severity Index. SAE International, 1971.
[403]Vollavanh L. R., K. M. O'Day, E. M. Koehling, J. M. May, K. M. Breedlove, E. L. Breedlove, . . . T. G. Bowman. Effect of impact mechanism on head accelerations in men's lacrosse athletes. *Journal of Applied Biomechanics* 34: 396-402, 2018.
[404]von Elm E., D. G. Altman, M. Egger, S. J. Pocock, P. C. Gøtzsche and J. P. Vandenbroucke. The Strengthening the Reporting of Observational Studies in Epidemiology (STROBE) statement: guidelines for reporting observational studies. *The Lancet* 370: 1453-1457, 2007.
[405]Weber A. F., J. P. Mihalik, J. K. Register-Mihalik, S. Mays, W. E. Prentice and K. M. Guskiewicz. Dehydration and performance on clinical concussion measures in collegiate wrestlers. *Journal of Athletic Training* 48: 153-160, 2013.
[406]Weik M. H. Nyquist theorem. In: *Computer Science and Communications Dictionary*. Boston, MA: Springer US, 2001, pp. 1127-1127.
[407]Wik E. H., O. Materne, K. Chamari, J. D. P. Duque, C. Horobeanu, B. Salcinovic, . . . A. Johnson. Involving research-invested clinicians in data collection affects injury incidence in youth football. *Scandinavian Journal of Medicine & Science in Sports* 2019.
[408]Wilcox B. J., J. G. Beckwith, R. M. Greenwald, J. J. Chu, T. W. McAllister, L. A. Flashman, . . . J. J. Crisco. Head impact exposure in male and female collegiate ice hockey players. *Journal of Biomechanics* 47: 109-114, 2014.
[409]Wilcox B. J., J. G. Beckwith, R. M. Greenwald, N. P. Raukar, J. J. Chu, T. W. McAllister, . . . J. J. Crisco. Biomechanics of head impacts associated with diagnosed concussion in female collegiate ice hockey players. *Journal of Biomechanics* 48: 2201-2204, 2015.
[410]Wilcox B. J., J. T. Machan, J. G. Beckwith, R. M. Greenwald, E. Burmeister and J. J. Crisco. Head-impact mechanisms in men's and women's collegiate ice hockey. *Journal of Athletic Training* 49: 514-520, 2014.
[411]Willigenburg N. W., J. R. Borchers, R. Quincy, C. C. Kaeding and T. E. Hewett. Comparison of injuries in American collegiate football and club rugby: a prospective cohort study. *American Journal of Sports Medicine* 44: 753-760, 2016.
[412]Willmott C., A. S. McIntosh, T. Howard, B. Mitra, B. Dimech-Betancourt, J. Donovan and J. V. Rosenfeld. SCAT3 changes from baseline and associations with X2 Patch measured head acceleration in amateur Australian football players. *Journal of Science and Medicine in Sport* 21: 442-446, 2017.
[413]Winkelstein B. A. and B. S. Myers. The biomechanics of cervical spine injury and implications for injury prevention. *Medicine & Science in Sports & Exercise* 29: S246-255, 1997.
[414]Wong R. H., A. K. Wong and J. E. Bailes. Frequency, magnitude, and distribution of head impacts in Pop Warner football: the cumulative burden. *Clinical Neurology and Neurosurgery* 118: 1-4, 2014.

[415]World Rugby. World Rugby's mission is "growing the global rugby family". 2016.

[416]Wu L. C., C. Kuo, J. Loza, M. Kurt, K. Laksari, L. Z. Yanez, . . . J. E. Urban. Detection of American football head impacts using biomechanical features and support vector machine classification. *Scientific Reports* 8: 855, 2017.

[417]Wu L. C., K. Laksari, C. Kuo, J. F. Luck, S. Kleiven, C. R. D. Bass and D. B. Camarillo. Bandwidth and sample rate requirements for wearable head impact sensors. *Journal of Biomechanics* 49: 2918-2924, 2016.

[418]Wu L. C., V. Nangia, K. Bui, B. Hammoor, M. Kurt, F. Hernandez, . . . D. B. Camarillo. In vivo evaluation of wearable head impact sensors. *Annals of Biomedical Engineering* 44: 1234-1245, 2016.

[419]Wu L. C., L. Zarnescu, V. Nangia, B. Cam and D. B. Camarillo. A head impact detection system using SVM classification and proximity sensing in an instrumented mouthguard. *IEEE Transactions on Bio-medical Engineering* 61: 2659-2668, 2014.

[420]Yeargin S. W., P. Kingsley, J. M. Mensch, J. P. Mihalik and E. V. Monsma. Anthropometrics and maturity status: A preliminary study of youth football head impact biomechanics. *International Journal of Psychophysiology* 132: 87-92, 2018.

[421]Yoganandan N. and F. A. Pintar. Biomechanics of human head-neck in rear impacts. *International Journal of Vehicle Design* 32: 53-67, 2003.

[422]Young T. J., R. W. Daniel, S. Rowson and S. M. Duma. Head impact exposure in youth football: elementary school ages 7-8 years and the effect of returning players. *Clinical Journal of Sport Medicine* 24: 416-421, 2014.

[423]Zhang L., K. H. Yang and A. I. King. A proposed injury threshold for mild traumatic brain injury. *Journal of Biomechanical Engineering* 126: 226-236, 2004.

[424]Zonner S. W., K. Ejima, Z. W. Bevilacqua, M. E. Huibregtse, C. Charleston, C. Fulgar and K. Kawata. Association of increased serum S100B levels with high school football subconcussive head impacts. *Frontiers in Neurology* 10: 327, 2019.

[425]Zuckerman S. L., Y. M. Lee, M. J. Odom, G. S. Solomon and A. K. Sills. Baseline neurocognitive scores in athletes with attention deficit–spectrum disorders and/or learning disability: Clinical article. *Journal of Neurosurgery. Pediatrics* 12: 103-109, 2013.

Appendices

Appendix A - Supplementary materials for Chapter 2

Methods

Incomplete information

Of the studies included in this review, three studies did not report the number of participants[111, 290, 349] and nine studies did not report the participant's sex.[20, 111, 114, 174, 210, 343, 349, 364, 422] All but three studies[343, 349, 364] reported on the age of the cohort, using an age group (e.g., collegiate), the mean age and SD or the age range. Nineteen studies[13, 37, 48, 78, 99, 147, 156, 188, 251, 270, 281, 284, 343, 349, 357, 359, 408, 409, 420] did not report any triggering or inclusion threshold and 13 studies[47, 107, 127, 153, 226, 251, 276, 297, 337, 343, 355, 356, 378, 416] utilized kinematics or injury metrics in their analyses but did not report summary values.

Cohorts

The cohorts identified are described in Table A.1.

Results

Head impact technology

The headgear-embedded HIT System (Simbex, Lebanon, NH, USA) was the first commercially available head impact telemetry system[115] and the most commonly utilised device (51% of included studies). The HIT System (Simbex, Lebanon, NH, USA) consisted of six spring-mounted single-axis accelerometers embedded in the helmet liner. It recorded linear accelerations and utilized proprietary algorithms to calculate the resultant peak linear accelerations at the head's centre of gravity. Because of the number of accelerometers, it is not possible to estimate the angular component about the vertical axis, therefore the HIT System is a 5-degree of freedom (5DOF) system and only provides an estimation of the angular acceleration at the moment of the peak linear acceleration.

Simbex developed another version to measure 6DOF accelerations, referred to as the 6DOF system. The 6DOF measurement device consisted of 12 single-axis accelerometers and provided the temporal angular acceleration response.[345] Four studies[116, 350, 351, 422] utilized both the HIT System and the 6DOF device. Since the 6DOF measurement device increased computation times[25] and the overall cost of the instrumented helmet[422] compared to the HIT System, it was discontinued and the last study reporting its use was published in 2014.[422]

The GForceTracker (GFT; GForceTracker Inc., Markham, Ontario, Canada) is a standalone device containing a triaxial linear accelerometer and a triaxial gyroscope. It was generally attached to the inner surface of a helmet but has also been used in a headband.

For the sports that do not entail the use of headgear, the skin-based xPatch (X2 Biosystems Inc., Seattle, WA, USA) device was the most popular devices, utilised in 21% of included studies. The xPatch was adhered to the skin behind the participant's ear over the mastoid process using different adhesives and methods of adhering. The xPatch contained a triaxial linear accelerometer and a triaxial gyroscope to capture 6DOF kinematics.

The Smart Impact Monitor (SIM or SIM-G, Triax Technologies Inc., Norwalk, CT) was placed in headbands or skull caps. The SIM-G also included a triaxial linear accelerometer and a triaxial gyroscope.

Five different mouthguards were also utilised: the Vector (Athlete Intelligence, Kirkland, WA, USA), the xGuard (X2Biosystems Inc., Seattle, WA, USA), the Prevent Impact Monitor (IM, or Intelligent Mouthguard, Prevent Biometrics, Edina, MN, USA), a custom mouthguard developed by Stanford University, and another custom mouthguard using sensors and battery components from the xPatch. One study did not report the type of skin patch utilised.[343]

Custom earplugs were equipped with either one triaxial accelerometer for the study of boxing impacts[364] or with three single-axis accelerometers in a pilot study reporting on rodeo.[245] In one study,[281] three single-axis accelerometers were arranged on headbands worn by football players. A single accelerometer was also implanted inside football and ice hockey helmets.[290]

Comparison of studies

The studies that were comparable were:

- [127, 194, 263, 398, 399] or [116, 194, 399] for the mean magnitude of impacts recorded with a 10 g triggering threshold (blue path on Figure A.1),
- [49, 59, 63, 81, 98, 397, 422] for the number of impacts with a 14.4 g triggering threshold (yellow path on Figure A.1),
- [59, 63, 81, 98, 397, 422] for the median PLA and PAA recorded with a 14.4 g triggering threshold (yellow path on Figure A.1).

Table A.1 Details of the cohorts that were reported on in multiple studies, as determined by in-text references, universities/teams and seasons, or comparison of number and age of participants.

Cohort	Sport	Age group	Technology	References
Individual cohorts being reported on multiple times				
Virginia Tech 2003-2010	Football	Collegiate	HIT System	50, 115, 116, 134, 135, 346, 350
University of North Carolina 2004-2012	Football	Collegiate	HIT System	149, 153, 157, 229, 263, 269, 304, 355
Dartmouth 2005-2011	Football and ice hockey	High school and collegiate	HIT System	151, 152, 249, 250
Illinois and Michigan high schools 2005-2015	Football	High school	HIT System	39, 41, 43, 45-49, 120, 244, 297
University of Rochester 2009-2014	Football	Collegiate	HIT System	23, 167, 261
University of North Carolina	Ice hockey	Youth and high school	HIT System	264-268, 271
Purdue University	Football	High school	HIT System	37, 78, 386
Wake Forest, High school, 2012	Football	High school	HIT System	99, 100, 397
Wake Forest, Youth, 2012-2015	Football	Youth	HIT System	13, 193, 194, 398, 399
Virginia Tech	Football	Youth	HIT System	59, 63
Temple University	Football	Collegiate	Vector mouthguard	190, 191
Stanford University, 2015	Football	Collegiate	Stanford mouthguard	220, 416
Irvington High School, 2017	Football	High school	Vector mouthguard	170, 424
Compilations of several cohorts				
NCAA and high school teams (no details) 2004-2006	Football	High school and collegiate	HIT System	147
NCAA and high school teams	Football	High school and collegiate	HIT System	26, 27, 46, 147, 149, 360
Brown University, Dartmouth College, and Virginia Tech, 2007-2010	Football and ice hockey	Collegiate	HIT System	90-92, 114, 248, 351
Brown, Dartmouth, 2008-2012	Ice hockey	Collegiate	HIT System	34, 408-410
NCAA-DoD CARE Consortium	Football	Collegiate	HIT System	64, 348, 352, 373-375
New Zealand rugby league	Rugby league	> 22 yr	xPatch	205, 208, 211

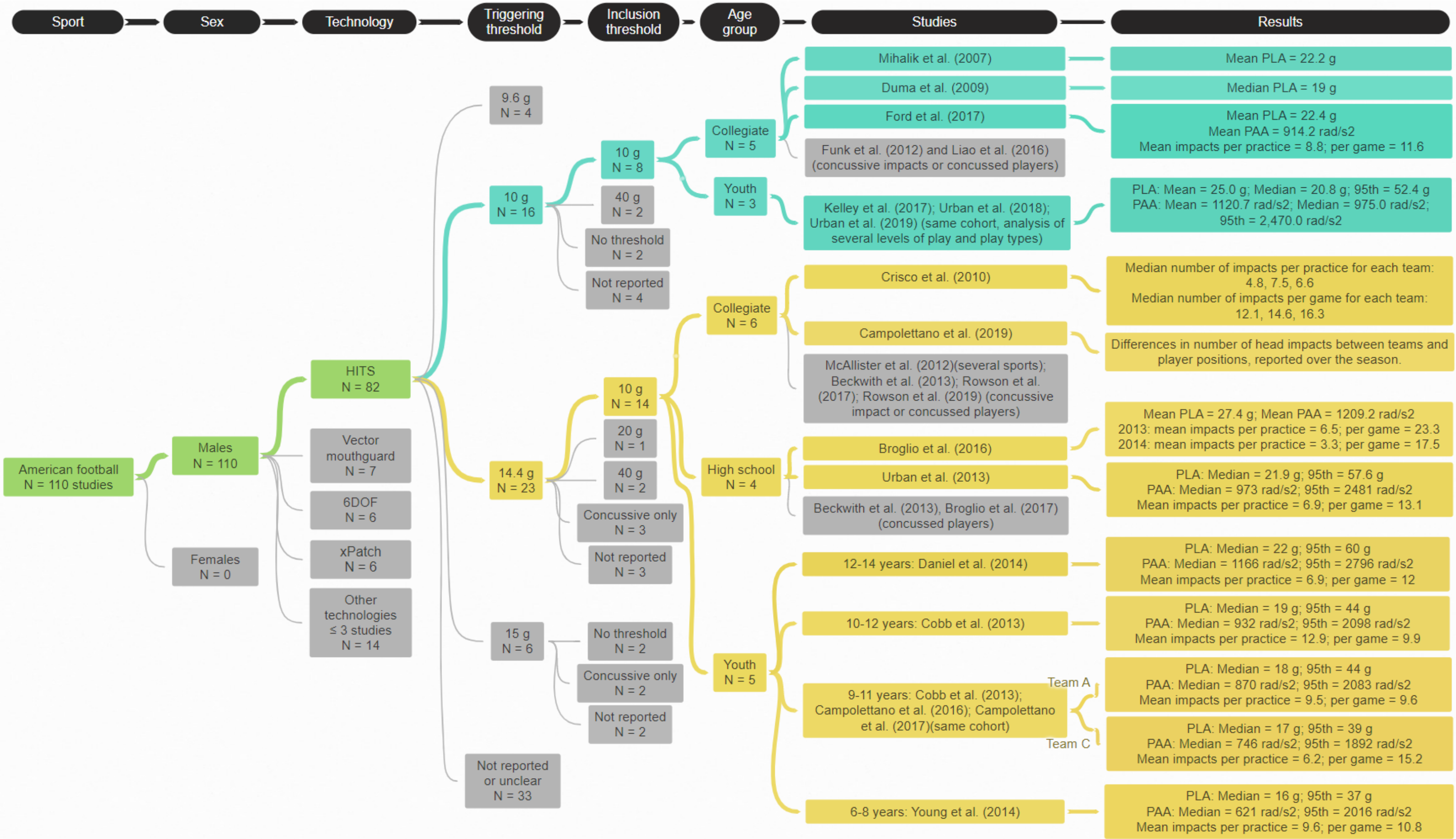


Figure A.1 Representation of 110 American football studies categorised by sex, technology used, triggering threshold, inclusion threshold and age group to illustrate the variability in methodologies. The blue and yellow paths with thick connecting lines represent the groupings with the most studies in each category, with the blue boxes corresponding to a 10 g-triggering threshold and the yellow boxes to a 14.4 g-triggering threshold. The grey boxes with thin connecting lines represent the categories with few studies, and the studies that were excluded from this analysis because of their focus on concussive impacts. HITS: Head Impact Telemetry System; 6DOF: 6-degree of freedom; PLA: Peak linear acceleration; PAA: Peak angular acceleration.

Appendix B - Supplementary materials for Chapter 3

Mouthguards axes orientation

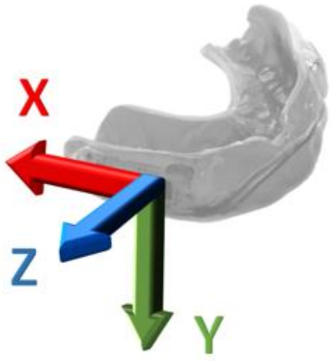


Figure B.1 Orientation of the CSx mouthguard's axes of measurement.

Angular velocity signal delayed onset

The angular velocity signal from CSx presented a delay with respect to the start of the LA pulse, of approximately 12 ms, confirmed by the manufacturer CSx (detailed in the chapter).

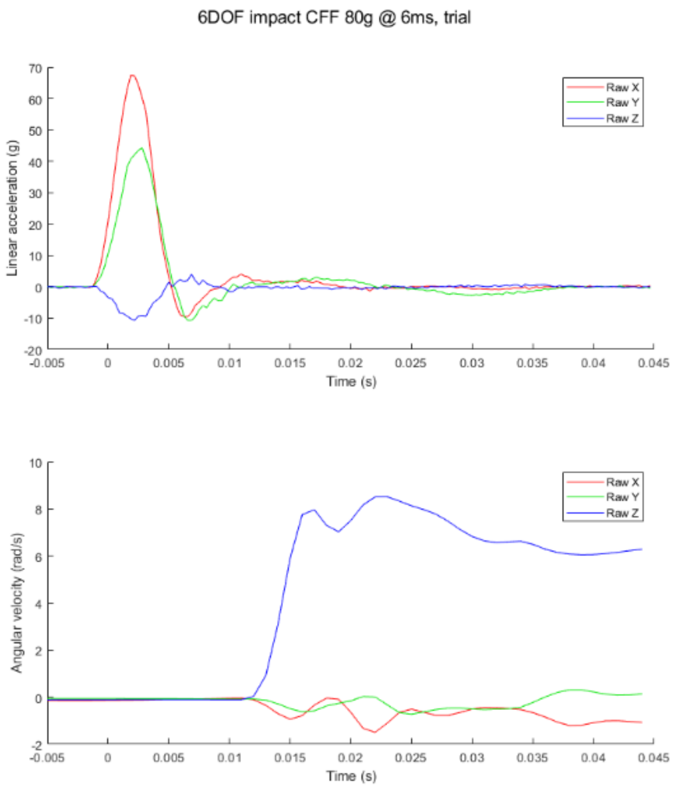


Figure B.2 Raw linear acceleration and angular velocity signals as measured by CSx for an example impact. The delay between the linear acceleration and the angular velocity is approximately 12 ms.

Linear impacts

Test setup

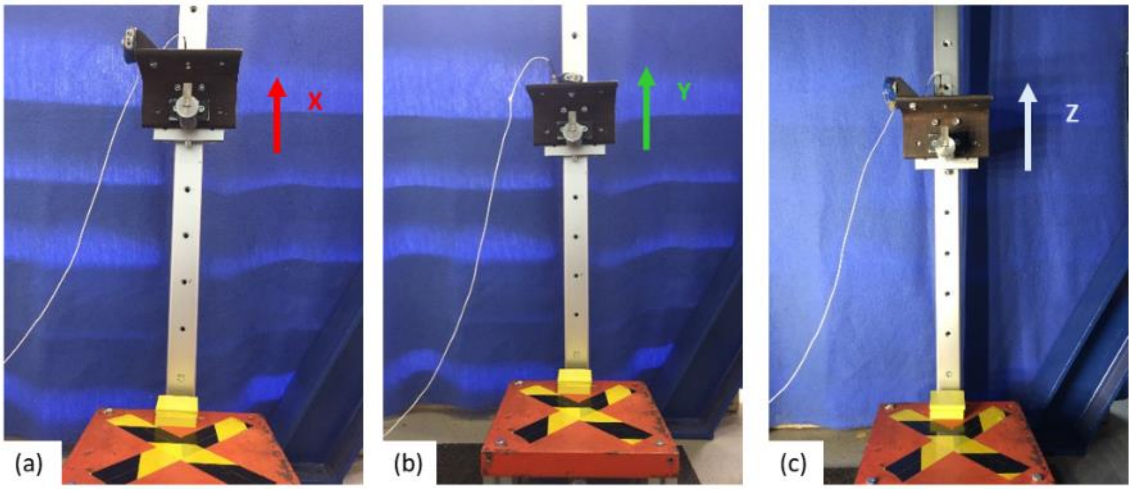


Figure B.3 One-degree-of-freedom linear test setup. (a) impact in the +X direction, (b) impact in the +Y direction, (c) impact in the +Z direction.

Test matrix

Table B.1 1DOF linear test matrix, with test conditions defined as the combination of targeted impact magnitude and duration.

Magnitude (g)	Duration (ms)	CSx sample ID X	Y	Z
40	15	28	10	10
40	20	28	10	28
80	6	28	10	10
80	15	28	10	10
130	6	28	10	10
170	6	28	10	10

Exclusions

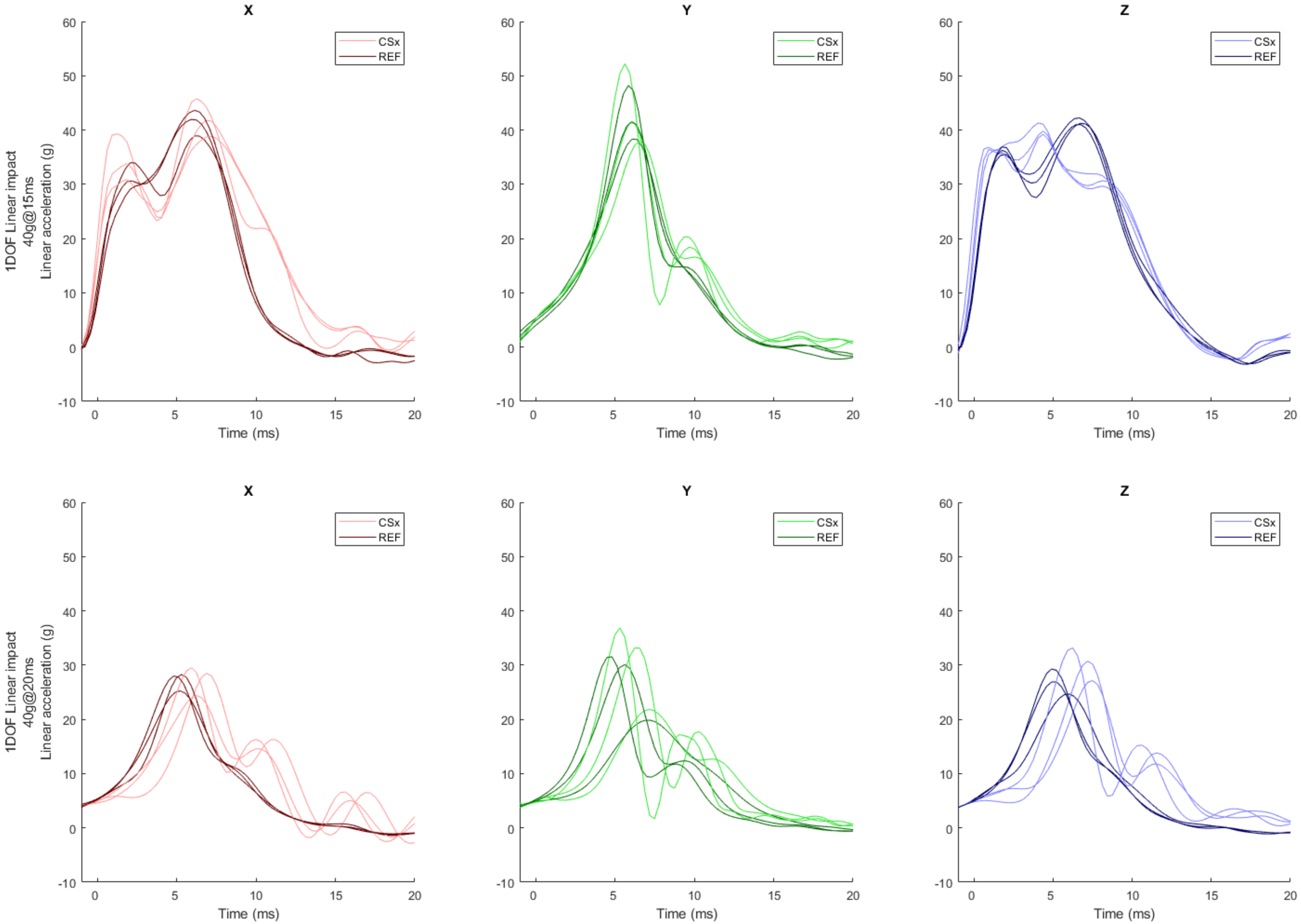
The CSx reached the accelerometer's saturation level (200 g) on six occasions during the 170g at 6ms linear impacts and such impacts were excluded from all analyses:

- 170g@6ms on X, trials 1 and 3
- 170g@6ms on Y, trials 1, 2, 3
- 170g@6ms on Z, trial 3

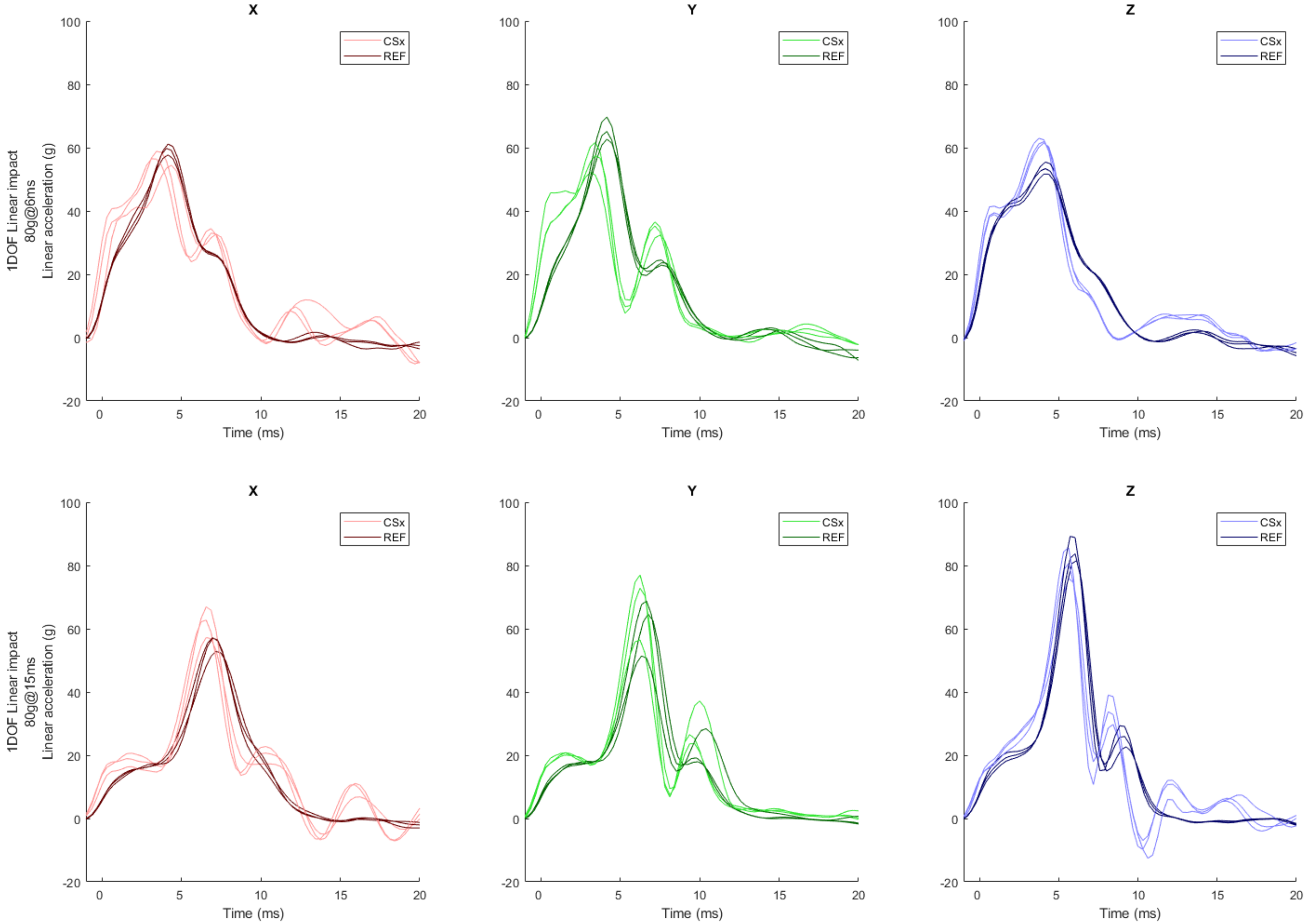
One additional trial along the Y-axis (130g@6ms on Y, trial 2) was excluded from the analysis of the area under the curve (AUC) and from CORA because the reference sensor recorded an abnormal spike close to the main region of interest (this did not affect the peak values).

All trials from the 40g@20ms conditions started with a resultant unfiltered CSx value over the chosen threshold used for aligning CSx and Ref signal, therefore it was not possible to align the signals using this method. It was chosen to exclude these trials for the AUC and CORA analyses.

Time series



X
Y
Z
CSx
REF
1DOF Linear impact
40g@15ms
Linear acceleration (g)
1DOF Linear impact
40g@20ms
Linear acceleration (g)
Time (ms)


X
Y
Z
CSx
REF
1DOF Linear impact
80g@6ms
Linear acceleration (g)
1DOF Linear impact
80g@15ms
Linear acceleration (g)
Time (ms)
100
80
60
40
20
0
-20
0
5
10
15
20

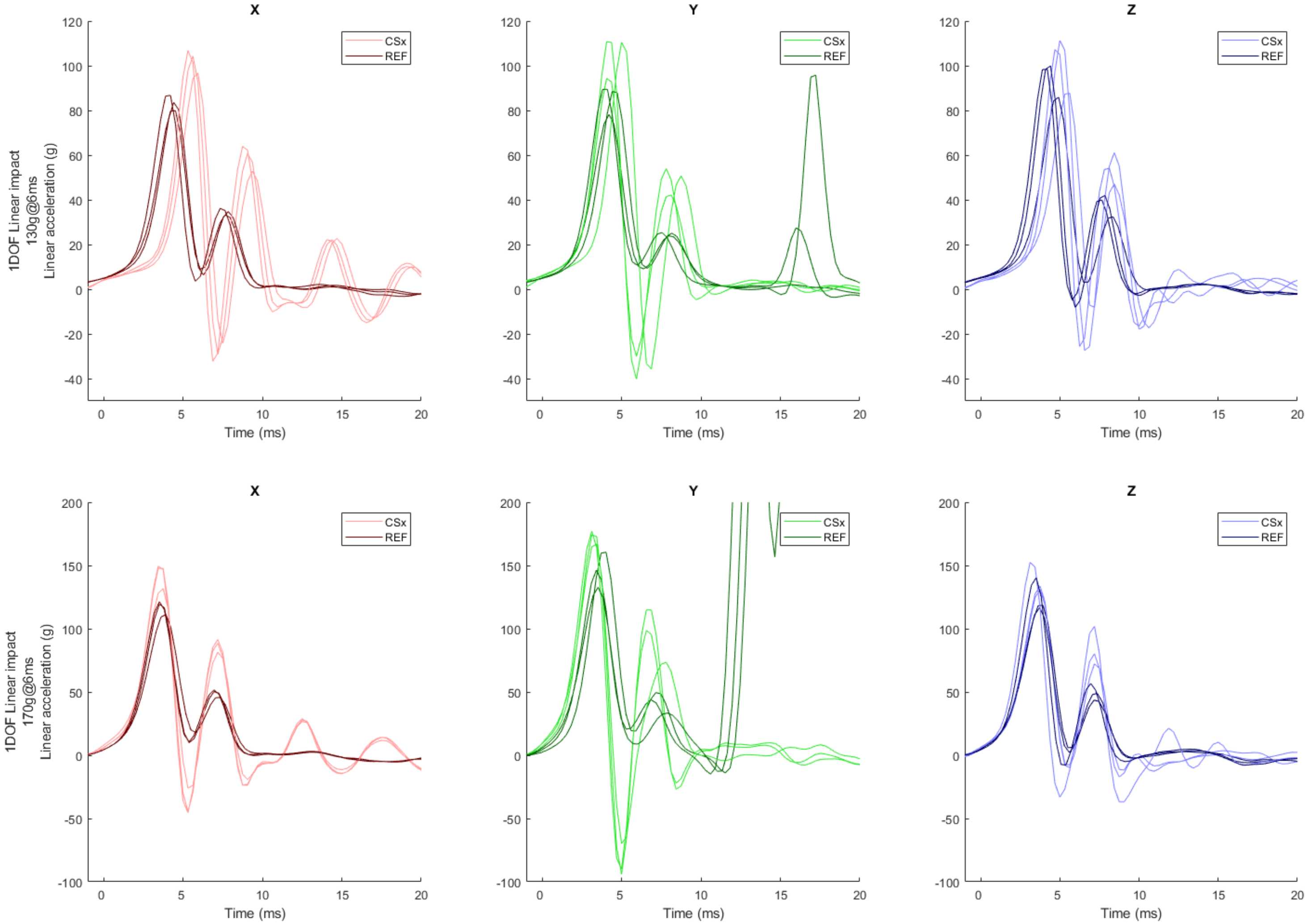


Figure B.4 CSx and REF linear acceleration time series from the 1DOF linear impacts.

Linear regressions

Peak

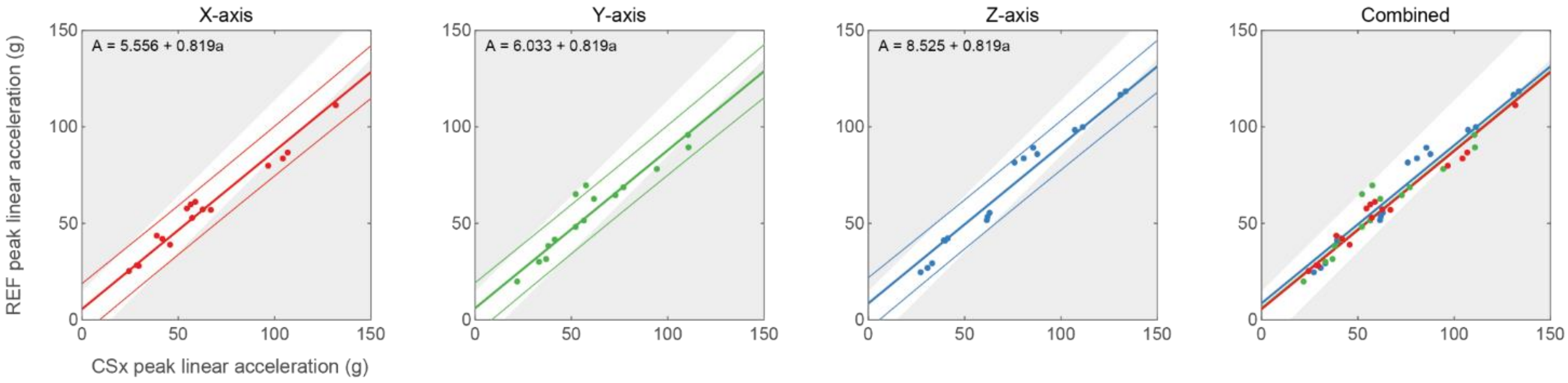


Figure B.5 Linear regression plots of the peak metric from the 1DOF linear impacts. The bold line represents the slope averaged across all directions of impacts; the fine lines represent the 95th percentile prediction interval; the shaded area represents ±10% of the plot's full scale about the diagonal.

Table B.2 Results of the robust linear regression models and error values for the peak linear acceleration from the 1DOF linear impacts.

	N	Estimate	SE	tStat	*p**	95th PI (% at the CSx mean)	MAE%	RMSE%
Average slope	48	0.819	0.029	6.252	<0.001	12.671(19%)		
Intercepts								
X	16	5.556	2.377	2.338	0.02		11%	14%
Y	15	6.033	2.196	0.217	**0.83**		12%	14%
Z	17	8.525	2.141	1.387	**0.17**		**9%**	11%

SE = standard error; 95th PI = half-width of the 95th percentile prediction interval; MAE% = Mean absolute error; RMSE% = Root mean square error.
*the p-value for the slope term indicates whether it is significantly different from 1, while the p-value for the intercept terms indicate whether it is significantly different from 0.

Area under the curve

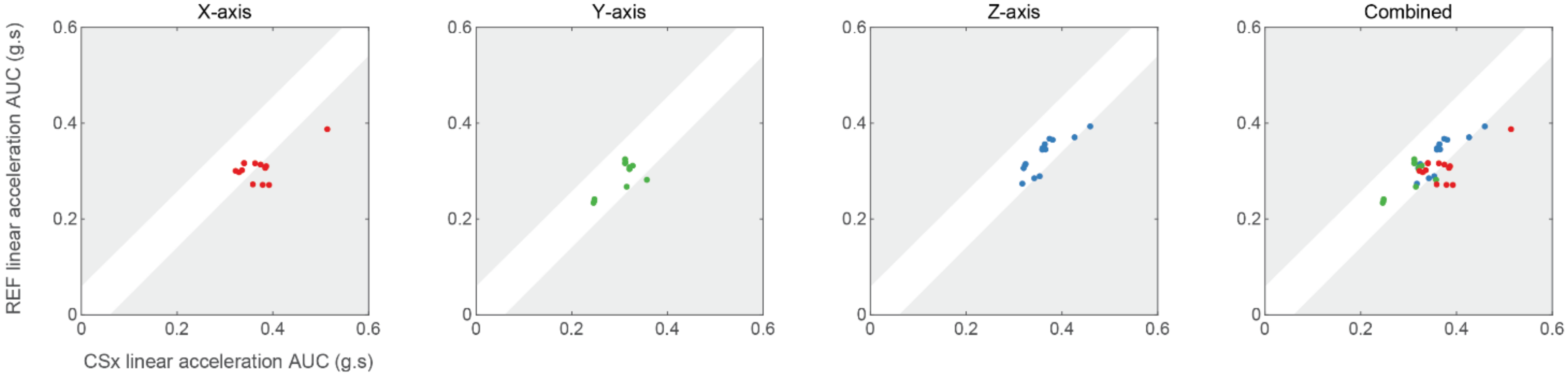


Figure B.6 Plots of the AUC metric from the 1DOF linear impacts. The limited spread of the data prevented the linear regression analysis. The shaded area represents ±10% of the plot's full scale about the diagonal.

Table B.3 Error values for the area under the linear acceleration curve from the 1DOF linear impacts.

Axis	N	MAE%	RMSE%
X	13	21%	25%
Y	11	7%	10%
Z	14	9%	11%

MAE% = Mean absolute error; RMSE% = Root mean square error.

CORA scores

Table B.4 Mean (standard deviation) and [range] of the overall CORA scores and sub-methods ratings for the 1DOF linear tests by axis of measurement.

		Overall CORA score	Corridor	Cross-correlation	Area under the curve	Time shift
	N	*Weights*	*0.50*	*0.167*	*0.167*	*0.167*
Linear tests, linear acceleration						
X	13	0.71 (0.12) [0.53 - 0.94]	0.64 (0.13) [0.49 - 0.89]	0.96 (0.02) [0.94 - 0.99]	0.68 (0.20) [0.39 - 0.97]	0.68 (0.20) [0.38 – 1.00]
Y	11	0.83 (0.08) [0.73 - 0.97]	0.80 (0.08) [0.69 - 0.95]	0.97 (0.03) [0.91 – 1.00]	0.80 (0.19) [0.43 - 0.98]	0.78 (0.21) [0.38 – 1.00]
Z	14	0.81 (0.05) [0.70 - 0.88]	0.75 (0.06) [0.65 - 0.89]	0.98 (0.01) [0.96 - 0.99]	0.82 (0.14) [0.60 - 0.98]	0.82 (0.13) [0.69 – 1.00]

Angular impacts

Test matrix

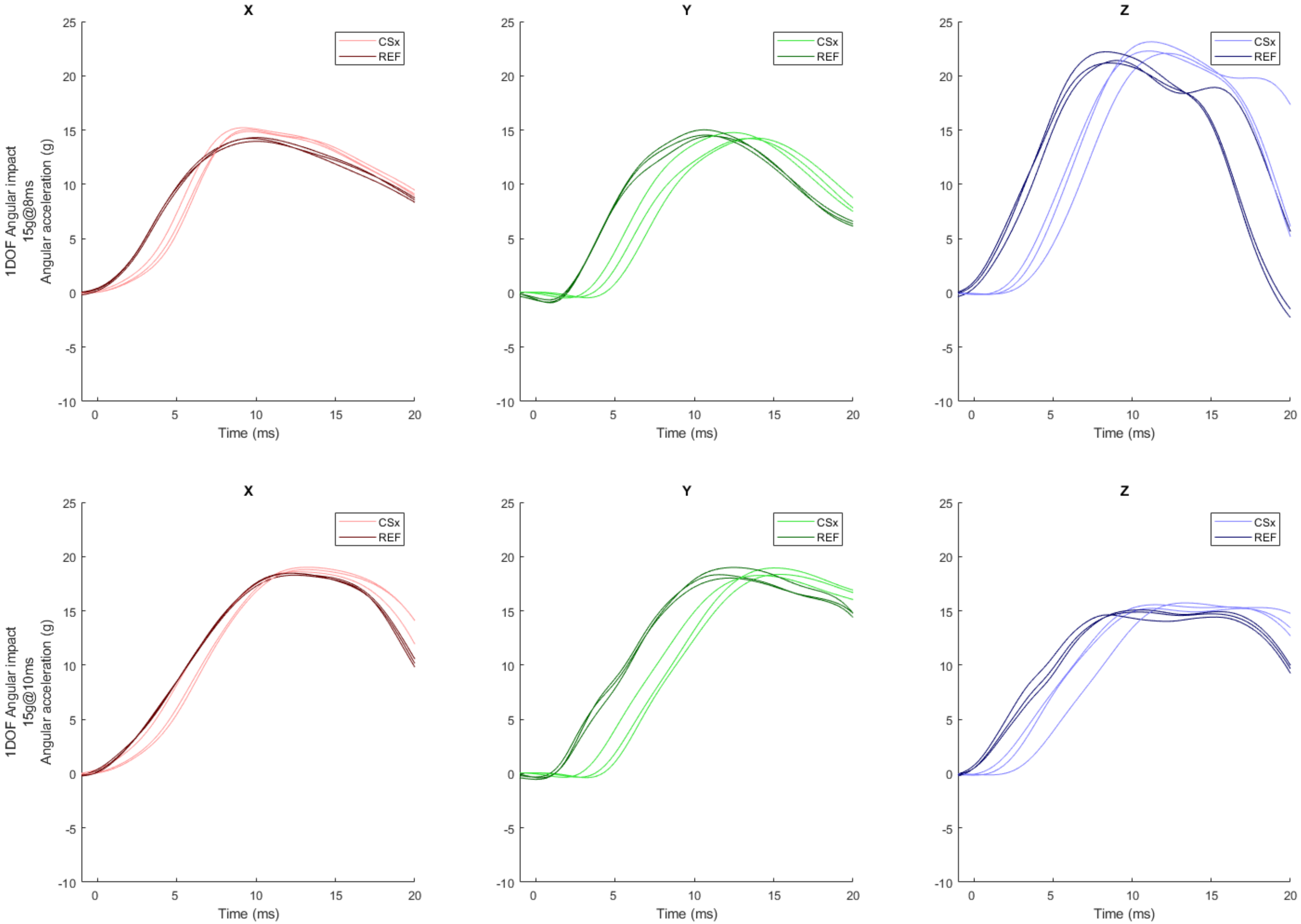
Table B.5 1DOF angular test matrix, with test conditions defined as the combination of targeted impact magnitude and duration.

Magnitude (rad/s)	Duration (ms)	CSx sample ID		
		X	Y	Z
15	8	31	28	28
15	10	31	28	28
25	8	31	28	28
25	10	31	28	31
40	8	31	28	28
40	10	31	28	31
40	12	31	28	31

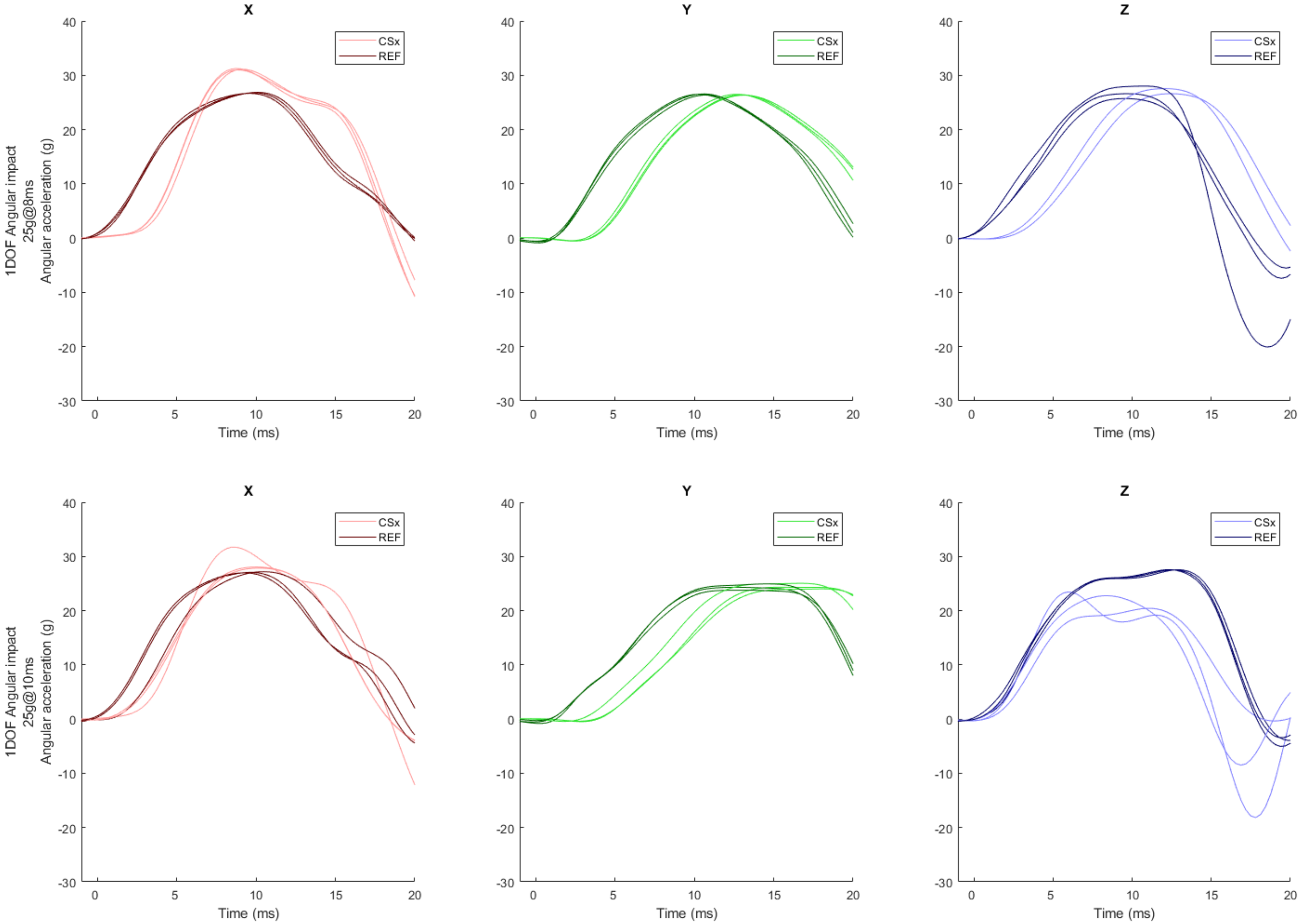
Exclusions

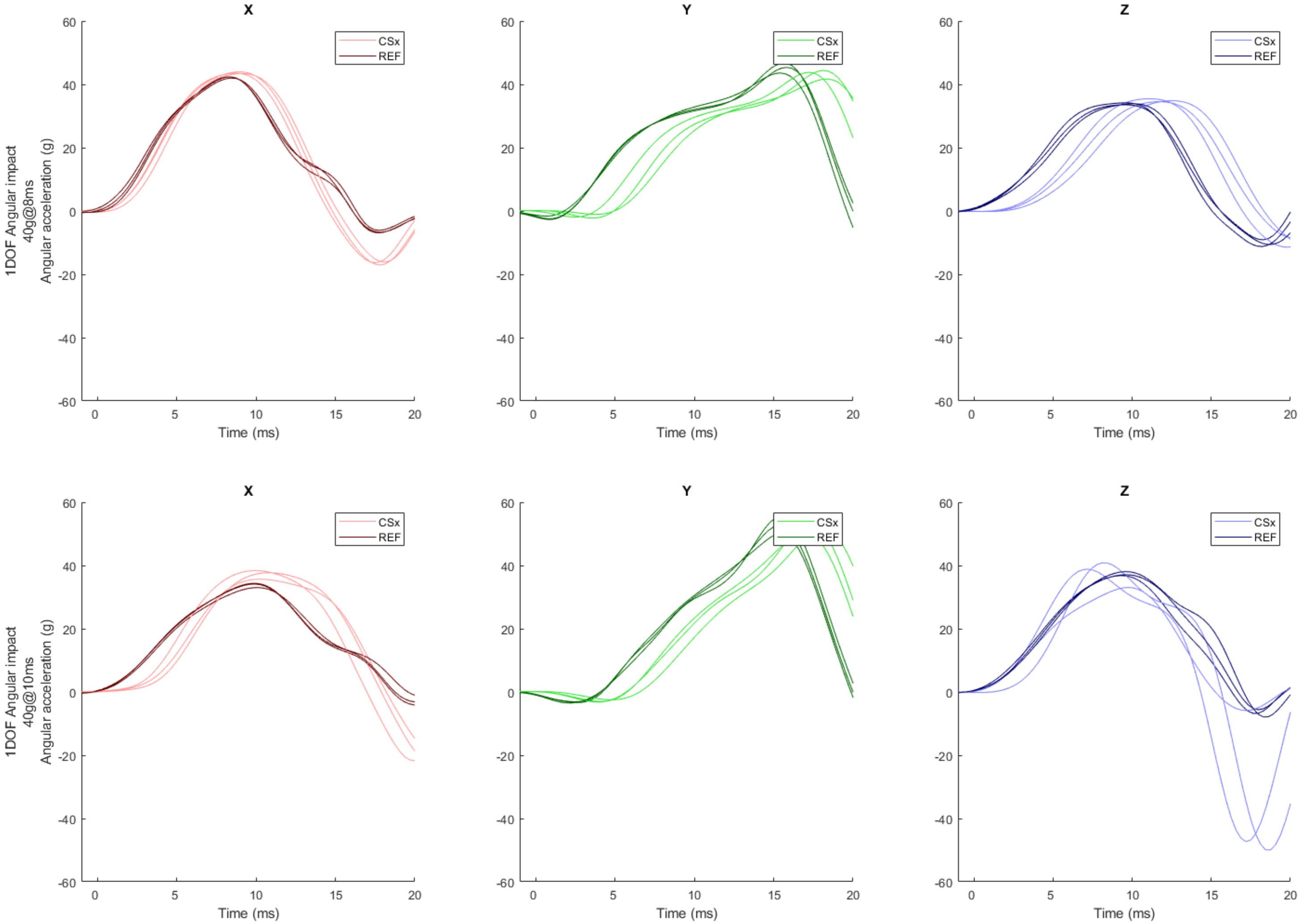
Two trials were excluded from all analyses based on the visual analysis of the time series (first trial of 25 rad.s^{-1} @ 10 ms about X, and third trial of 40 rad.s^{-1} @ 10 ms about Z). In both cases, the excluded trial appeared substantially different from the other two repeated trials. These may have been motion artefacts from the mouthguard.

Time series



X
Y
Z
CSx
REF
1DOF Angular impact
15g@8ms
15g@10ms
Angular acceleration (g)
Time (ms)
25
20
15
10
5
0
-5
-10


X
Y
Z
CSx
REF
1DOF Angular impact
25g@8ms
Angular acceleration (g)
1DOF Angular impact
25g@10ms
Angular acceleration (g)
Time (ms)
40
30
20
10
0
-10
-20
-30
0
5
10
15
20


X
Y
Z
CSx
REF
1DOF Angular impact
40g@8ms
Angular acceleration (g)
1DOF Angular impact
40g@10ms
Angular acceleration (g)
Time (ms)
60
40
20
0
-20
-40
-60
0
5
10
15
20

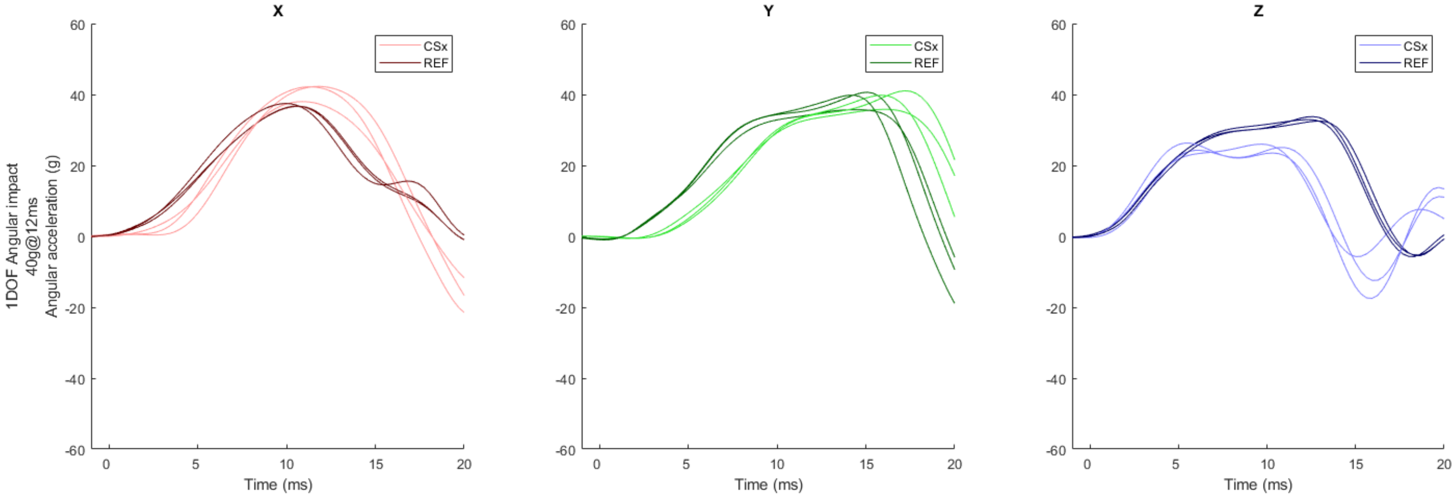


Figure B.7 CSx and REF angular velocity time series from the 1DOF angular impacts.

Linear regressions

Peak

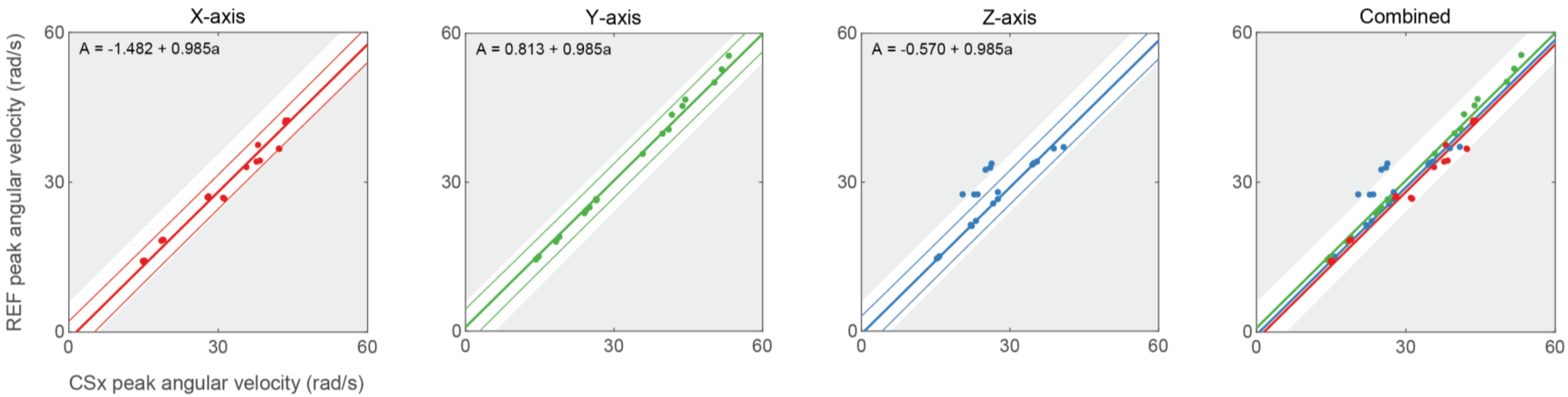


Figure B.8 Linear regression plots of the peak metric from the 1DOF angular impacts. The bold line represents the slope averaged across all directions of impacts; the fine lines represent the 95th percentile prediction interval; the shaded area represents ±10% of the plot's full scale about the diagonal.

Table B.6 Results of the robust linear regression models and error values for the peak angular velocity from the 1DOF linear impacts.

	N	Estimate	SE	tStat	p*	95th PI (% at the CSx mean)	MAE%	RMSE%
Average slope	61	0.985	0.020	0.744	**0.460**	3.423 (12%)		
Intercepts								
X	20	-1.482	0.733	-2.022	0.048		**8%**	9%
Y	21	0.813	0.521	4.408	<0.001		**1%**	2%
Z	20	-0.570	0.535	1.704	**0.094**		**9%**	12%

SE = standard error; 95th PI = half-width of the 95th percentile prediction interval; MAE% = Mean absolute error; RMSE% = Root mean square error.
*the p-value for the slope term indicates whether it is significantly different from 1, while the p-value for the intercept terms indicate whether it is significantly different from 0.

Area under the curve

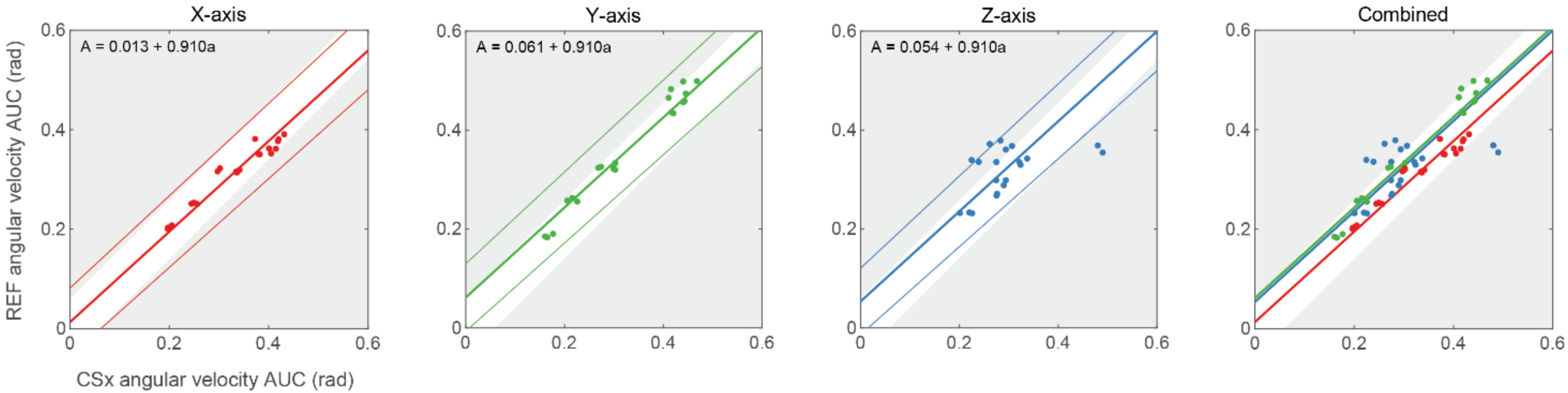


Figure B.9 Linear regression plots of the AUC metric from the 1DOF angular impacts. The bold line represents the slope averaged across all directions of impacts; the fine lines represent the 95th percentile prediction interval; the shaded area represents ±10% of the plot's full scale about the diagonal.

Table B.7 Results of the robust linear regression models and error values for the area under the angular velocity curve from the 1DOF angular impacts.

	N	Estimate	SE	tStat	p*	95th PI (% at the CSx mean)	MAE%	RMSE%
Average slope	61	0.910	0.045	2.008	0.049	0.062 (20%)		
Intercepts								
X	20	0.013	0.016	0.810	0.421		7%	8%
Y	21	0.061	0.009	5.116	<0.001		10%	11%
Z	20	0.054	0.010	4.195	<0.001		14%	19%

SE = standard error; 95th PI = half-width of the 95th percentile prediction interval; MAE% = Mean absolute error; RMSE% = Root mean square error. *the p-value for the slope term indicates whether it is significantly different from 1, while the p-value for the intercept terms indicate whether it is significantly different from 0.

CORA scores

Table B.8 Mean (standard deviation) and [range] of the overall CORA scores and sub-methods ratings for the 1DOF angular tests by axis of measurement.

	N	Overall CORA score	Corridor	Cross-correlation	Area under the curve	Time shift
		Weights	*0.50*	*0.167*	*0.167*	*0.167*
Angular tests, angular velocity						
X	20	0.74 (0.11) [0.57 - 0.96]	0.68 (0.16) [0.47 - 0.98]	0.98 (0.02) [0.95 – 1.00]	0.75 (0.15) [0.52 – 1.00]	0.68 (0.20) [0.40 – 1.00]
Y	21	0.68 (0.05) [0.61 - 0.78]	0.57 (0.09) [0.44 - 0.74]	0.99 (0.01) [0.97 – 1.00]	0.88 (0.08) [0.74 – 1.00]	0.47 (0.06) [0.40 - 0.52]
Z	20	0.59 (0.09) [0.45 - 0.80]	0.48 (0.12) [0.33 - 0.80]	0.95 (0.05) [0.82 – 1.00]	0.66 (0.28) [0.24 - 0.98]	0.48 (0.22) [0.15 – 1.00]

6DOF impacts

Test matrix

Table B.9 6DOF test matrix.

Flat tests		CSX sample ID	
Magnitude (g)	Duration (ms)	Flat frontal	Flat lateral
40	6	31	
40	12	32	32
40	16	32	32
80	6	31	32
80	12	31	32
80	16	31	-
130	6	32	32
130	12	32	32
130	16	-	32
170	6	31	32
170	12	-	32

Oblique tests Magnitude (rad.s^{-1})	Duration (ms)	Oblique coronal	Oblique sagittal	Oblique horizontal
5	10	32	-	-
5	17	32	31	32
5	25	32	31	32
15	10	32	31	32
15	17	32	31	32
15	25	-	31	32
25	10	32	31	32
25	17	32	31	32

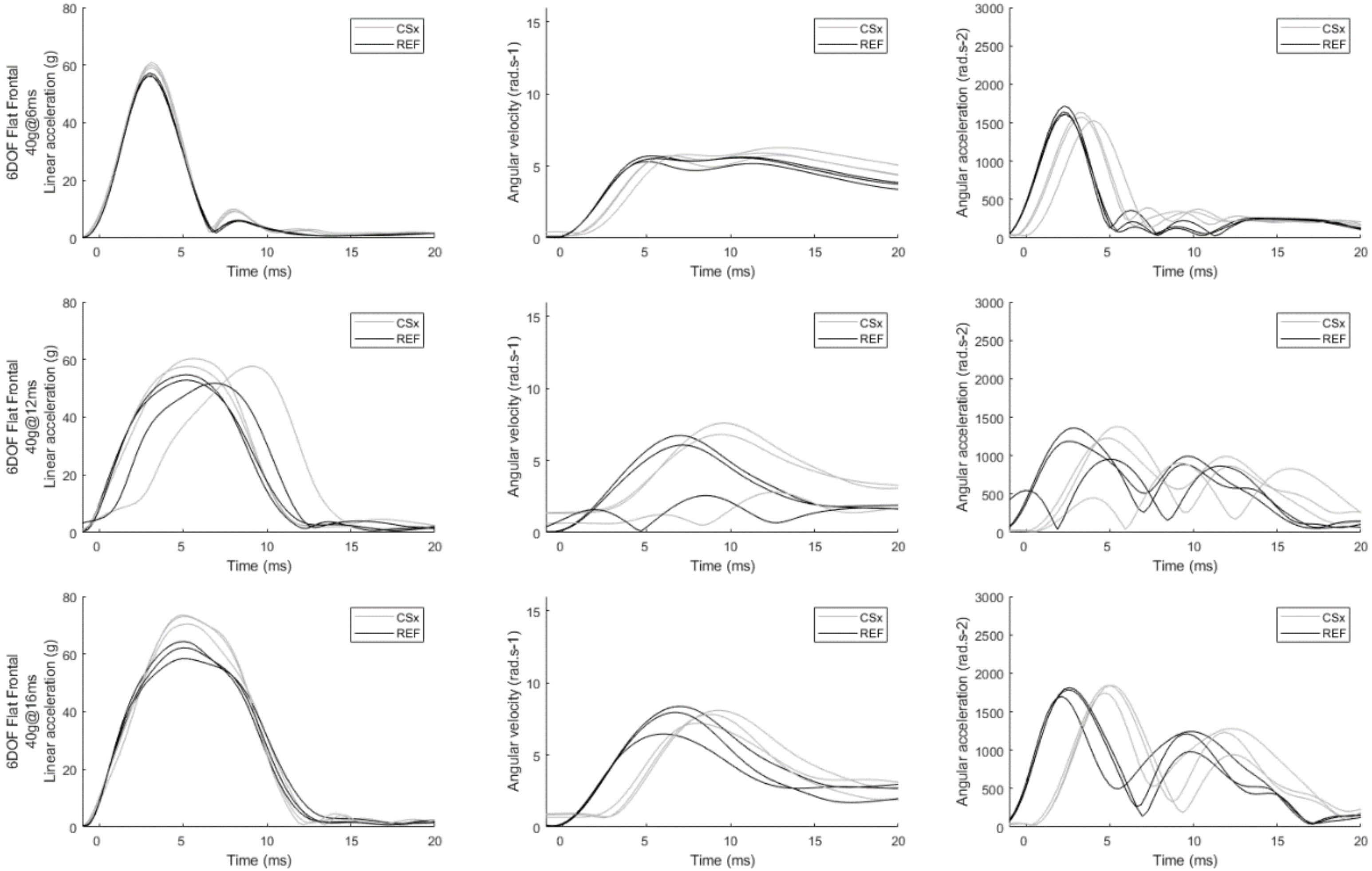
Exclusions

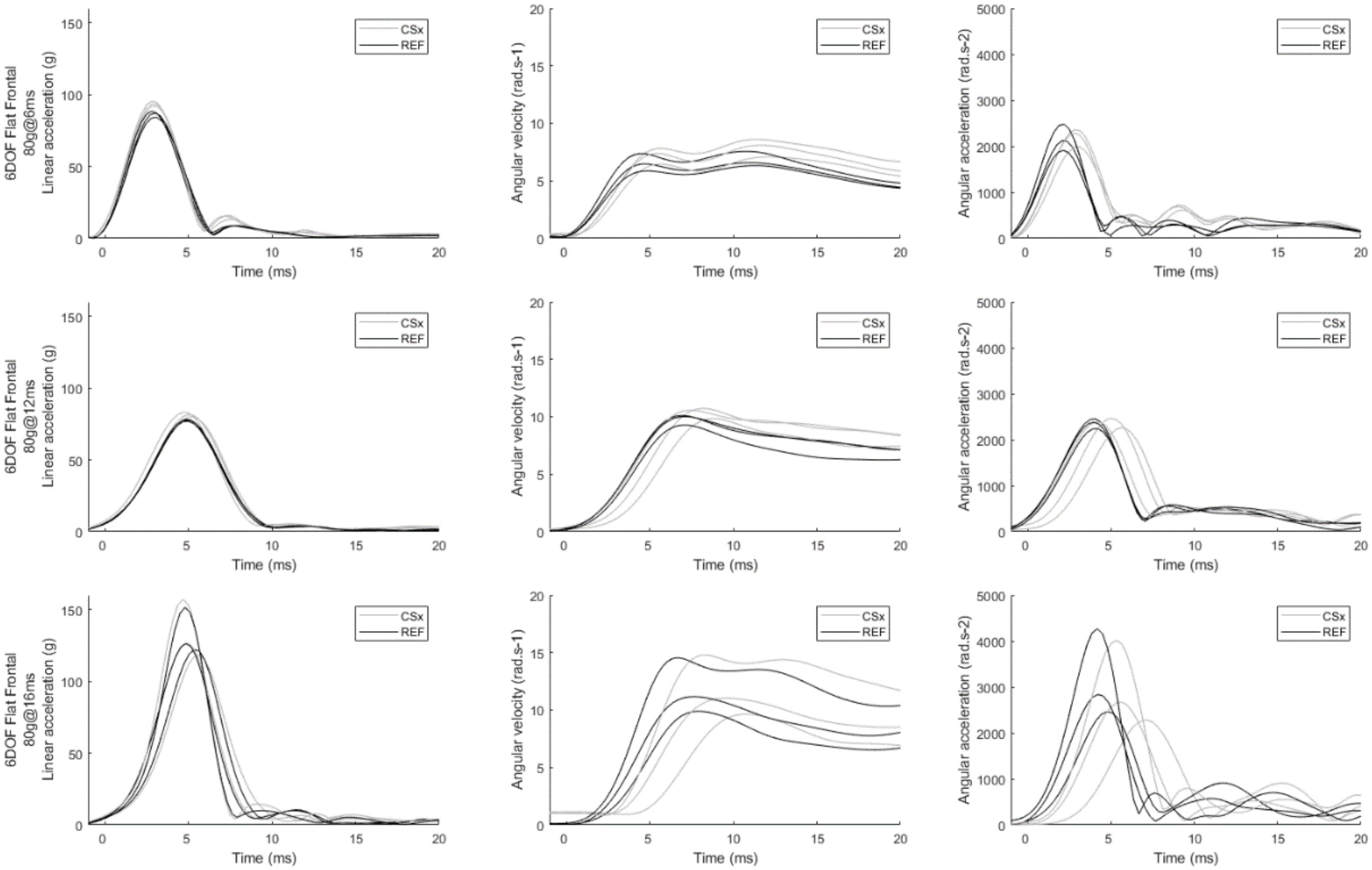
One trial (CFF 40@12_3) was excluded from all analyses as both CSx and REF showed a substantially different curve from the other two trials of the same condition.

Two trials from the COX 5 rad.s^{-1} @17ms (2nd and 3rd), and all trials from COY 5 rad.s^{-1} @25ms started with a CSx value over the chosen threshold used for aligning CSx and Ref signal, therefore it was not possible to align the signals using this method. It was chosen to exclude these trials for the AUC and CORA analyses.

Time series

Flat frontal




6DOF Flat Frontal
80g@6ms
Linear acceleration (g)
Angular velocity (rad.s-1)
Angular acceleration (rad.s-2)
6DOF Flat Frontal
80g@12ms
6DOF Flat Frontal
80g@16ms
Time (ms)
CSx
REF

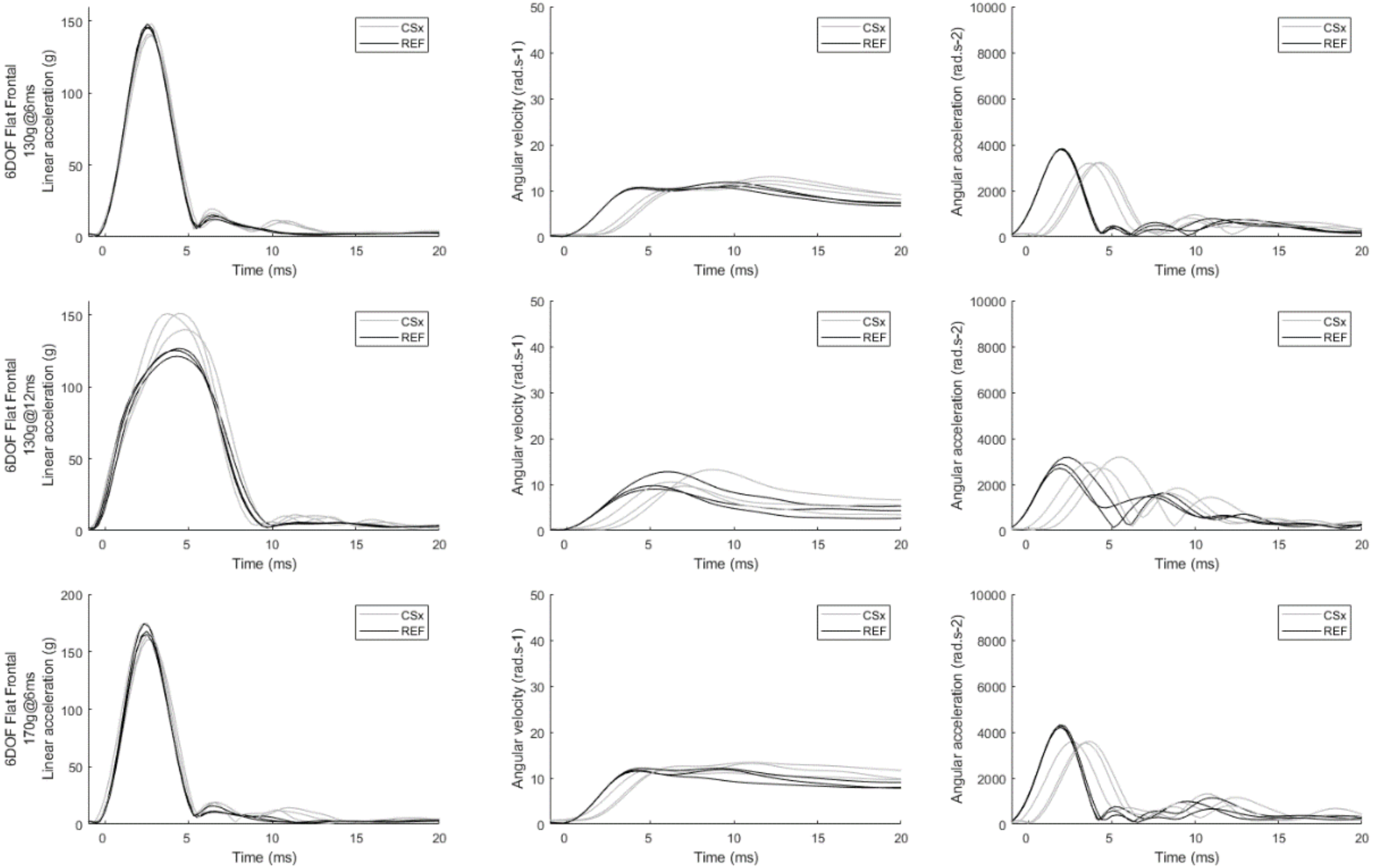
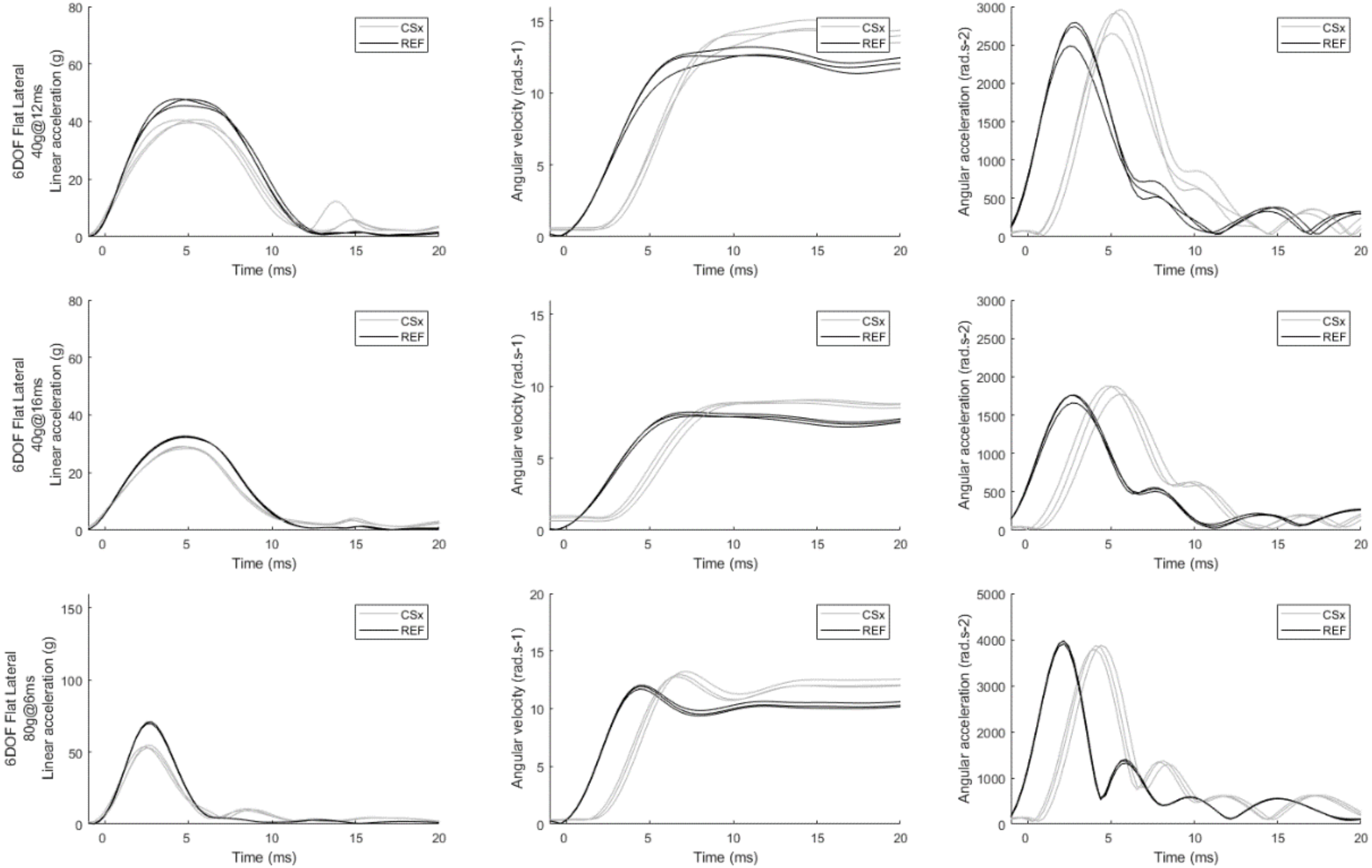
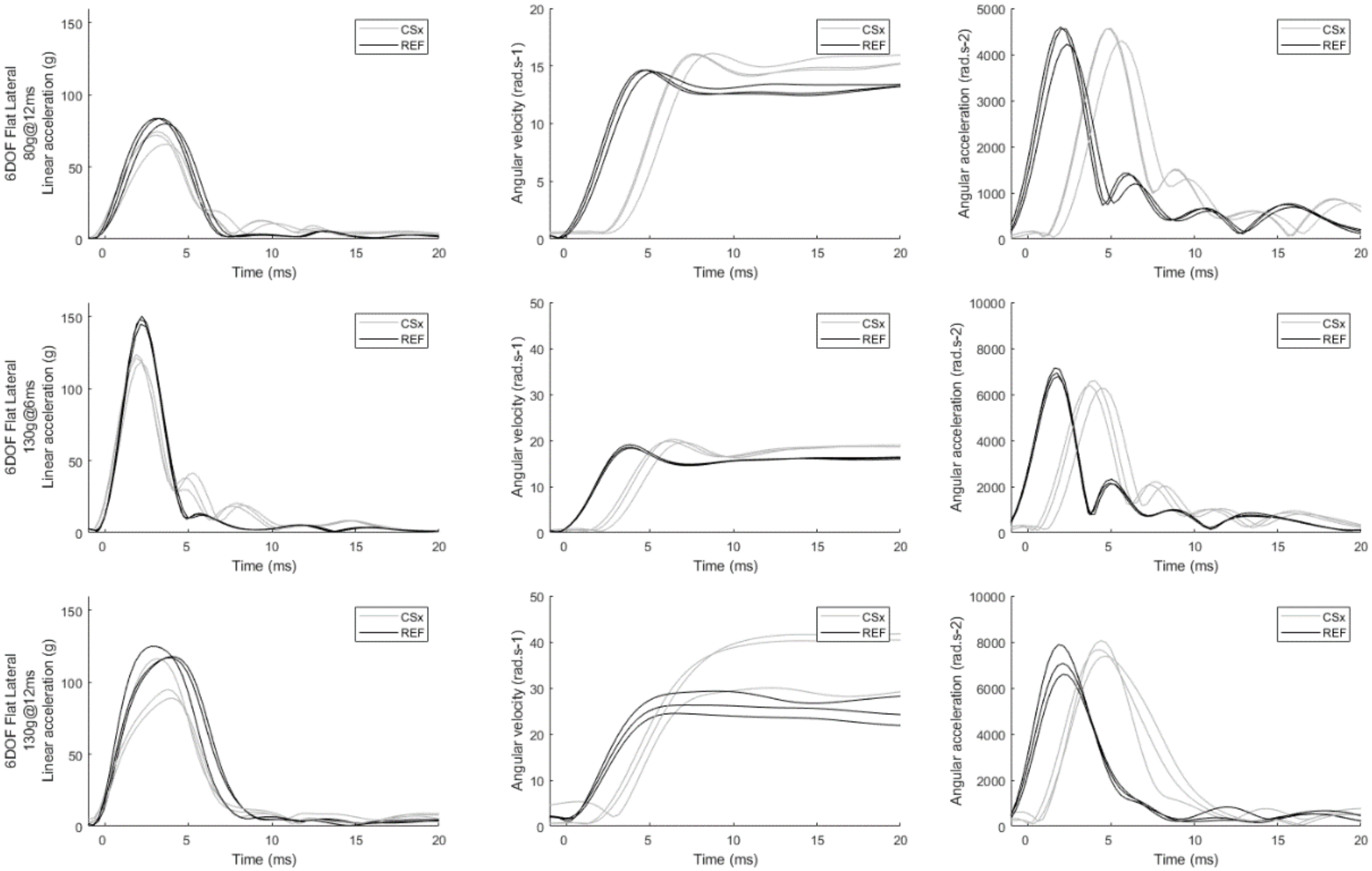


Figure B.10 CSx and REF linear acceleration, angular velocity and angular acceleration time series from the 6DOF Flat frontal impacts.

Flat lateral





6DOF Flat Lateral
80g@12ms
Linear acceleration (g)
Angular velocity (rad.s-1)
Angular acceleration (rad.s-2)
6DOF Flat Lateral
130g@6ms
Linear acceleration (g)
6DOF Flat Lateral
130g@12ms
Linear acceleration (g)
Time (ms)
CSx
REF

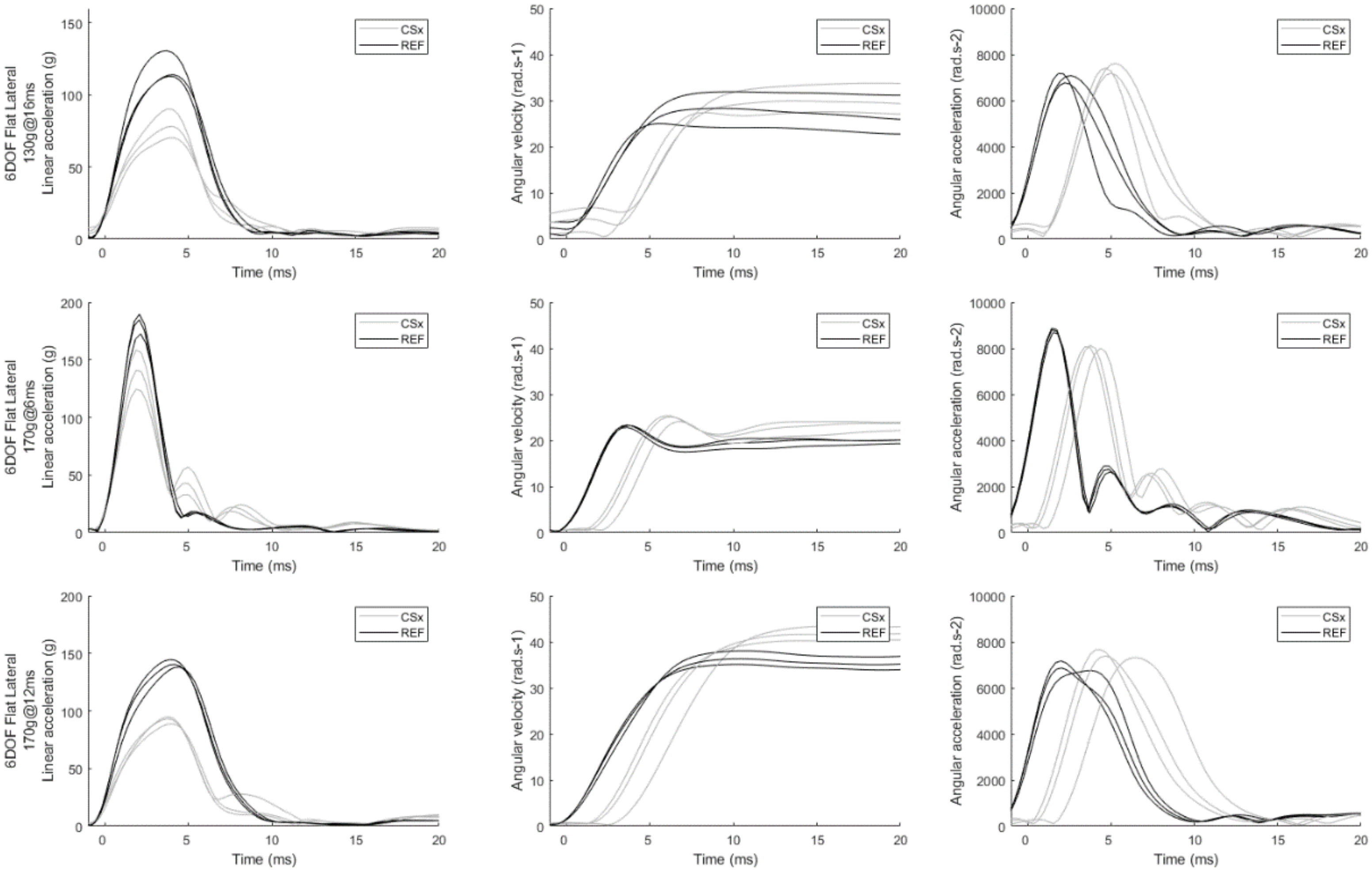
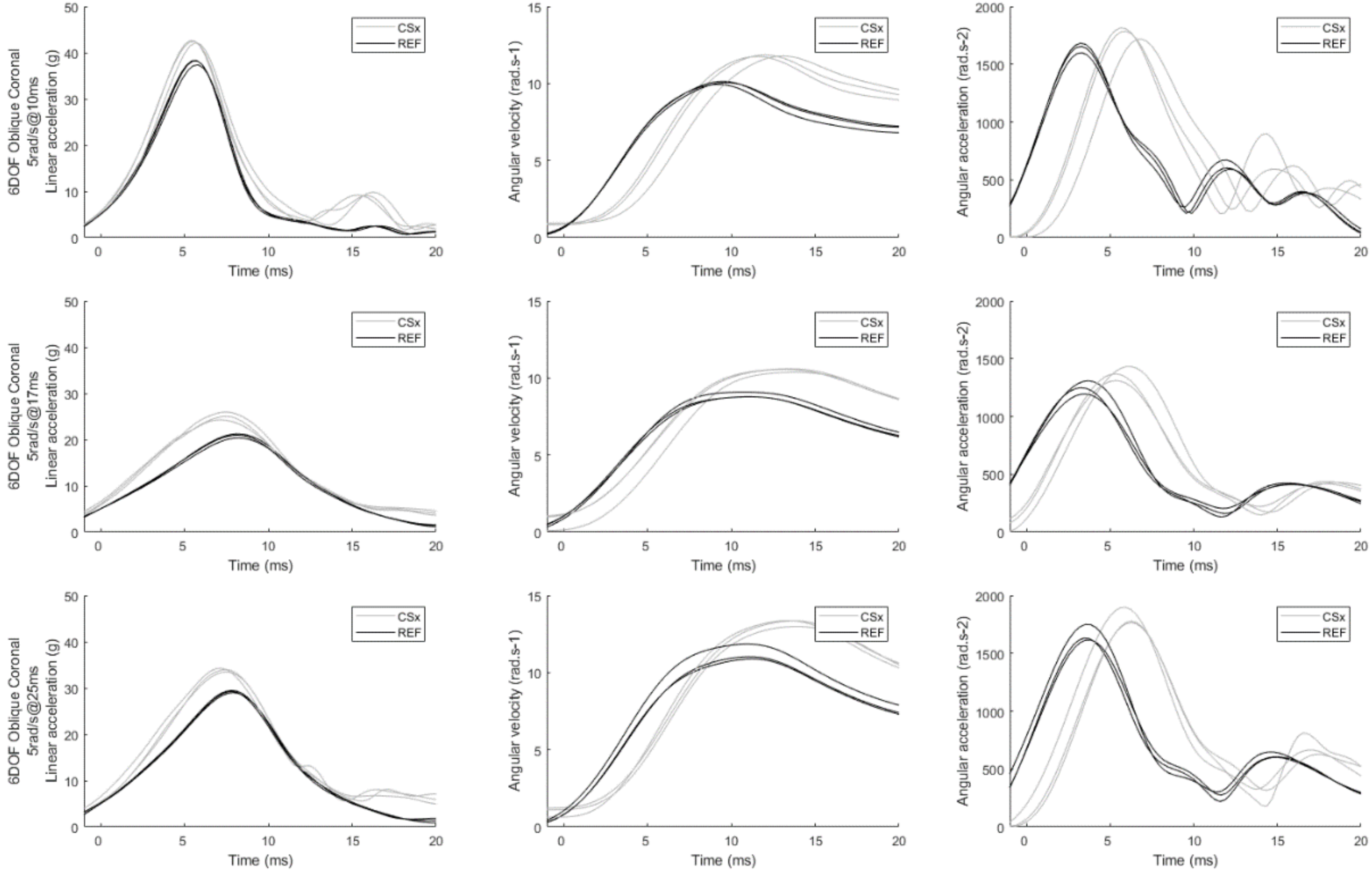
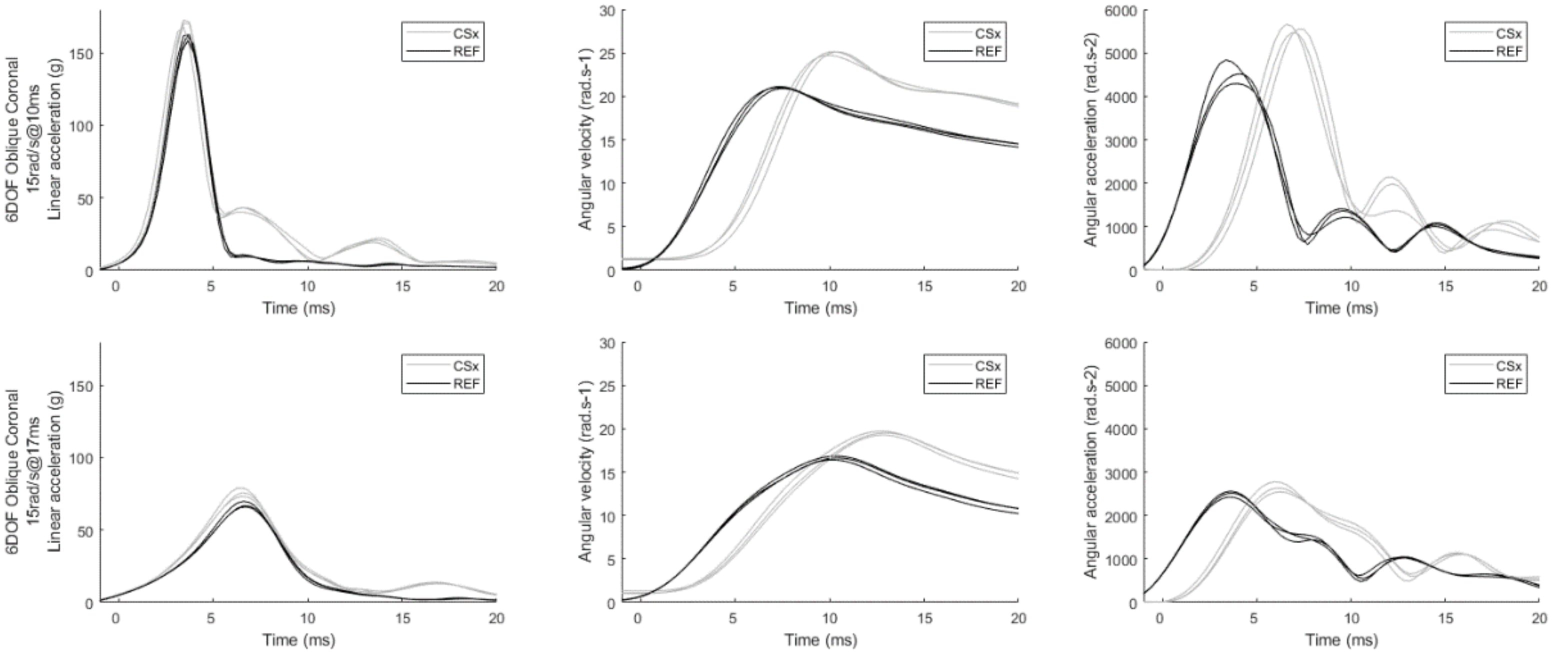


Figure B.11 CSx and REF linear acceleration, angular velocity and angular acceleration time series from the 6DOF Flat lateral impacts.

Oblique coronal




6DOF Oblique Coronal
15rad/s@10ms
Linear acceleration (g)
Angular velocity (rad.s-1)
Angular acceleration (rad.s-2)
Time (ms)
CSx
REF
6DOF Oblique Coronal
15rad/s@17ms
Linear acceleration (g)
Angular velocity (rad.s-1)
Angular acceleration (rad.s-2)
Time (ms)
CSx
REF

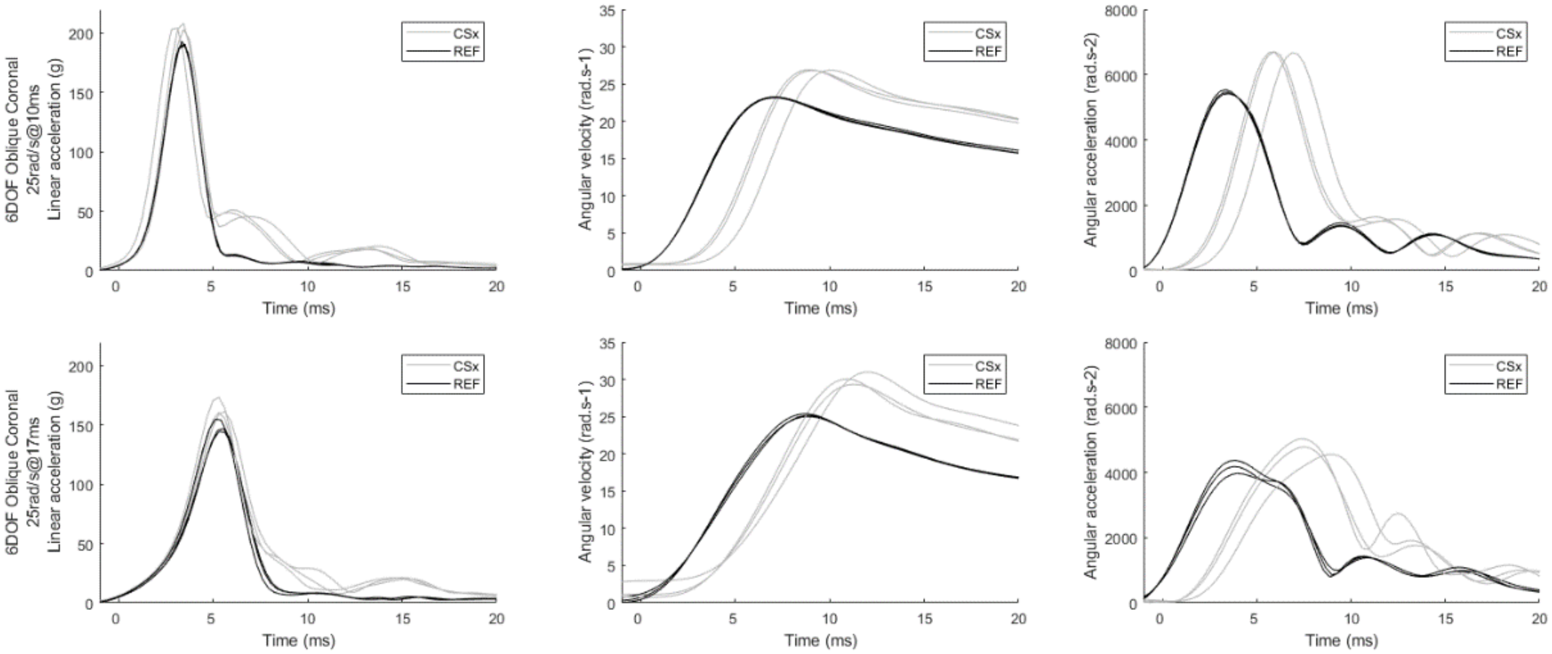
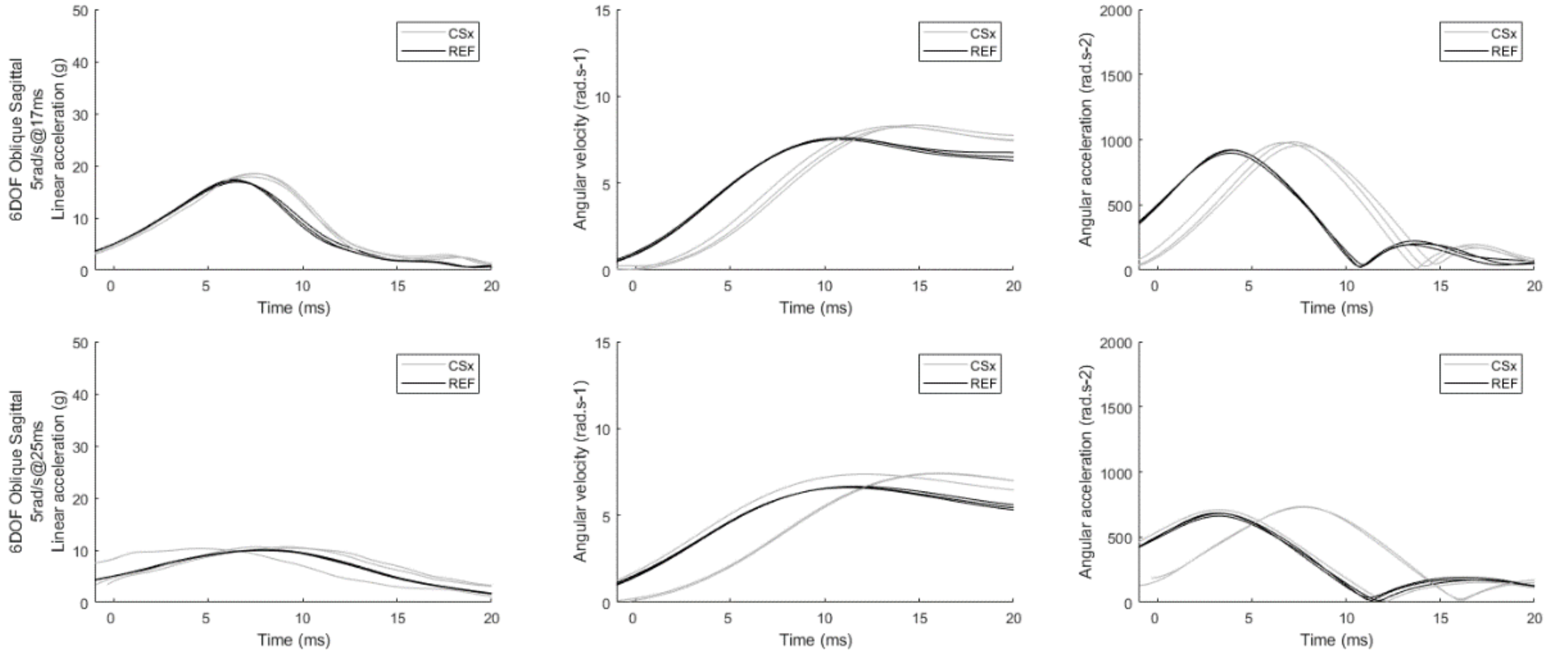
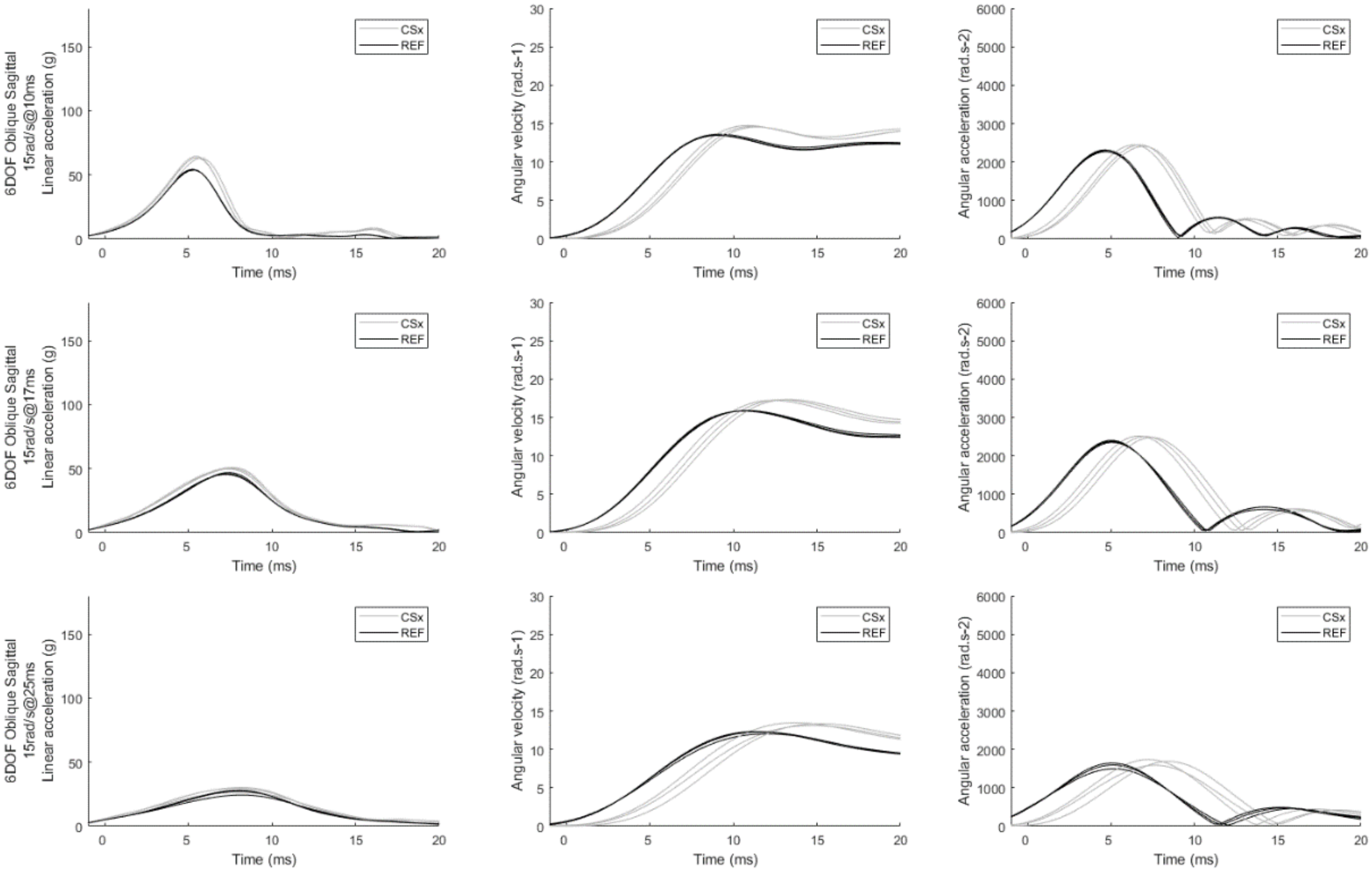


Figure B.12 CSx and REF linear acceleration, angular velocity and angular acceleration time series from the 6DOF Oblique coronal impacts.

Oblique sagittal





6DOF Oblique Sagittal
15rad/s@10ms
Linear acceleration (g)
Angular velocity (rad.s-1)
Angular acceleration (rad.s-2)
CSx
REF
Time (ms)
6DOF Oblique Sagittal
15rad/s@17ms
6DOF Oblique Sagittal
15rad/s@25ms

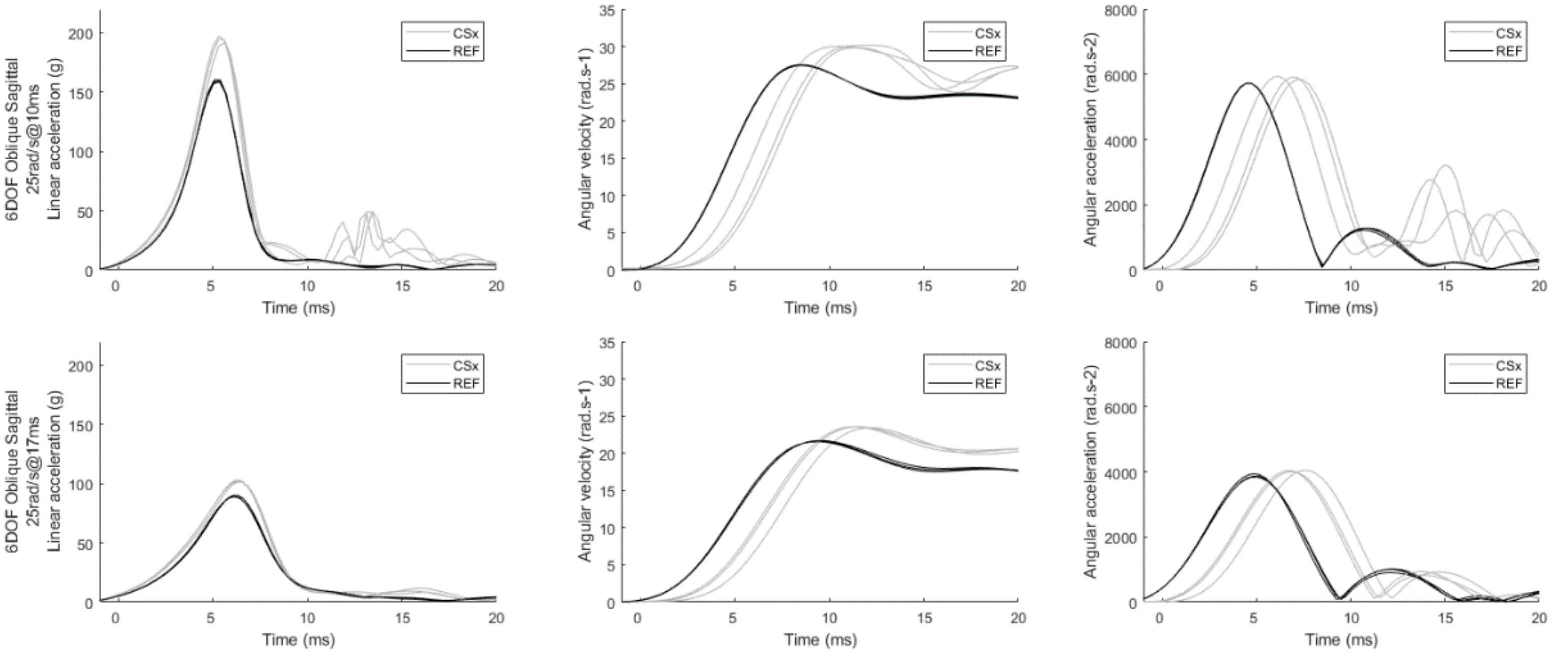
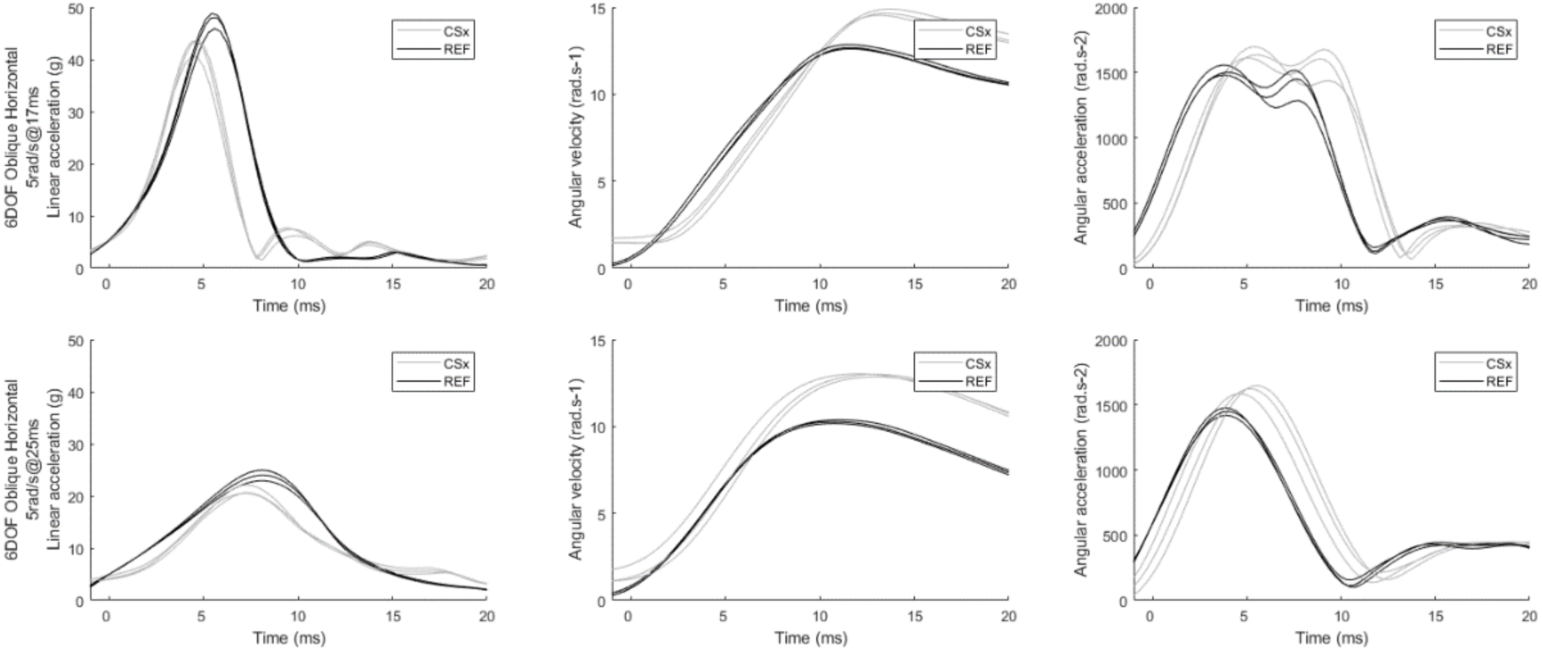
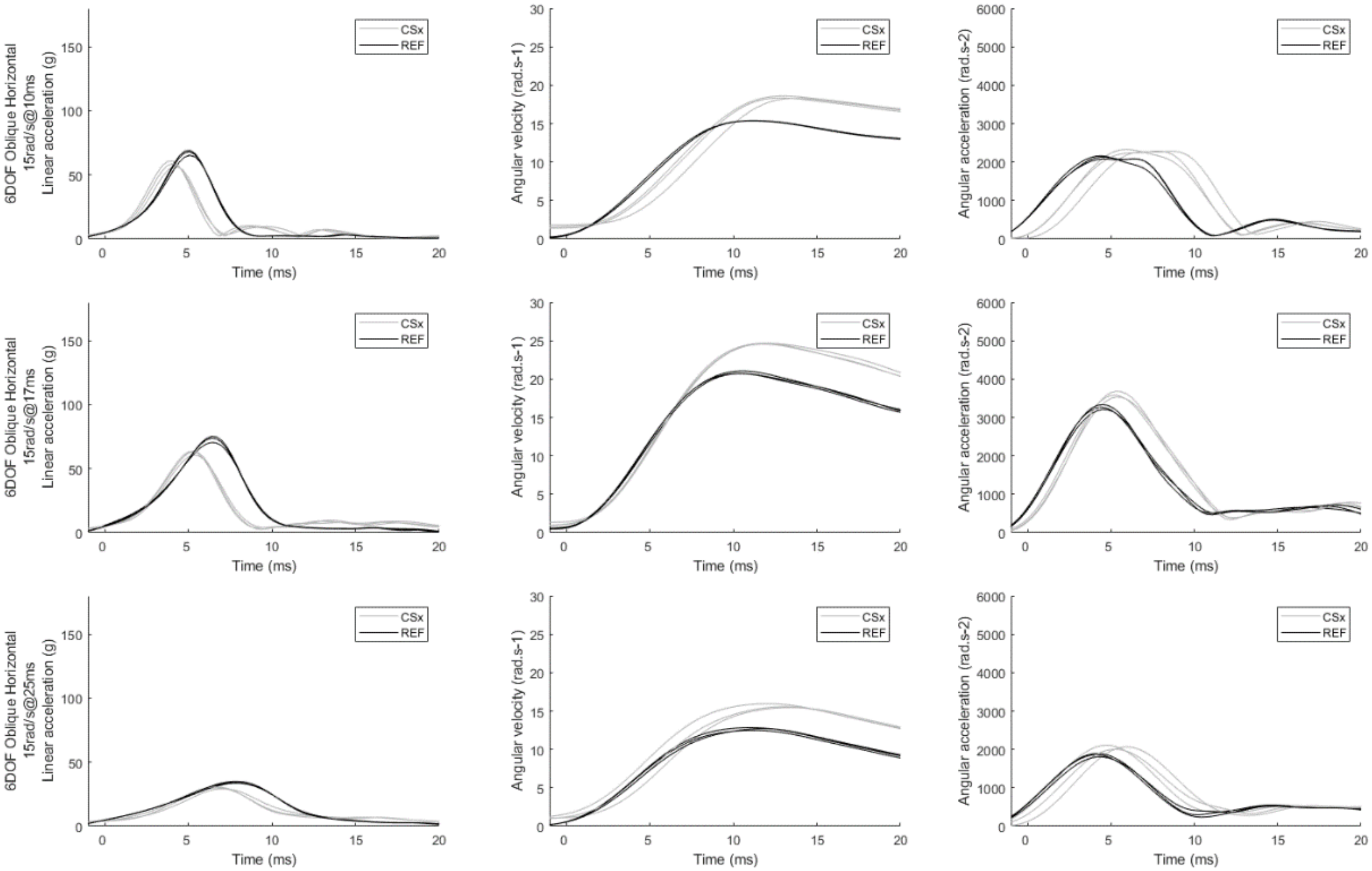


Figure B.13 CSx and REF linear acceleration, angular velocity and angular acceleration time series from the 6DOF Oblique sagittal impacts.

Oblique horizontal




6DOF Oblique Horizontal
15rad/s@10ms
Linear acceleration (g)
Angular velocity (rad.s-1)
Angular acceleration (rad.s-2)
CSx
REF
Time (ms)
6DOF Oblique Horizontal
15rad/s@17ms
Linear acceleration (g)
6DOF Oblique Horizontal
15rad/s@25ms
Linear acceleration (g)

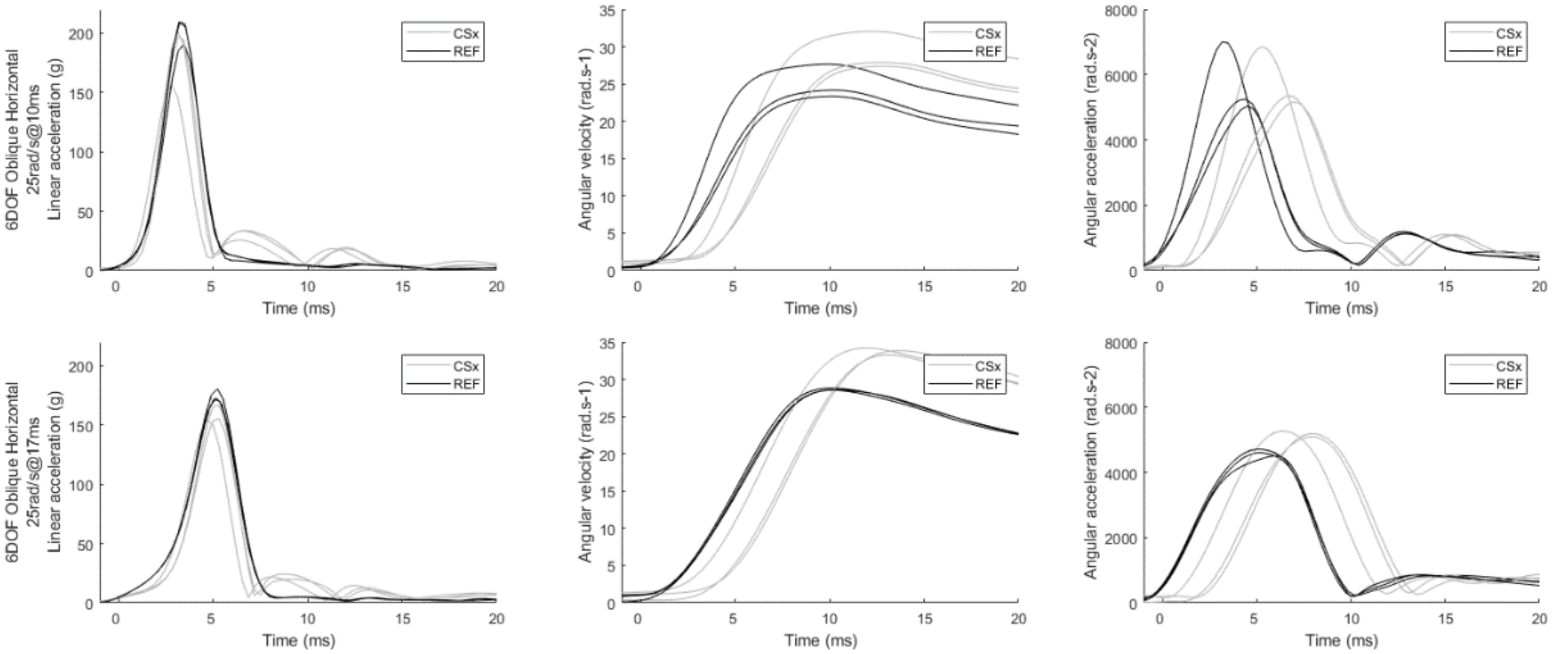


Figure B.14 CSx and REF linear acceleration, angular velocity and angular acceleration time series from the 6DOF Oblique horizontal impacts.

AUC Linear regression and errors

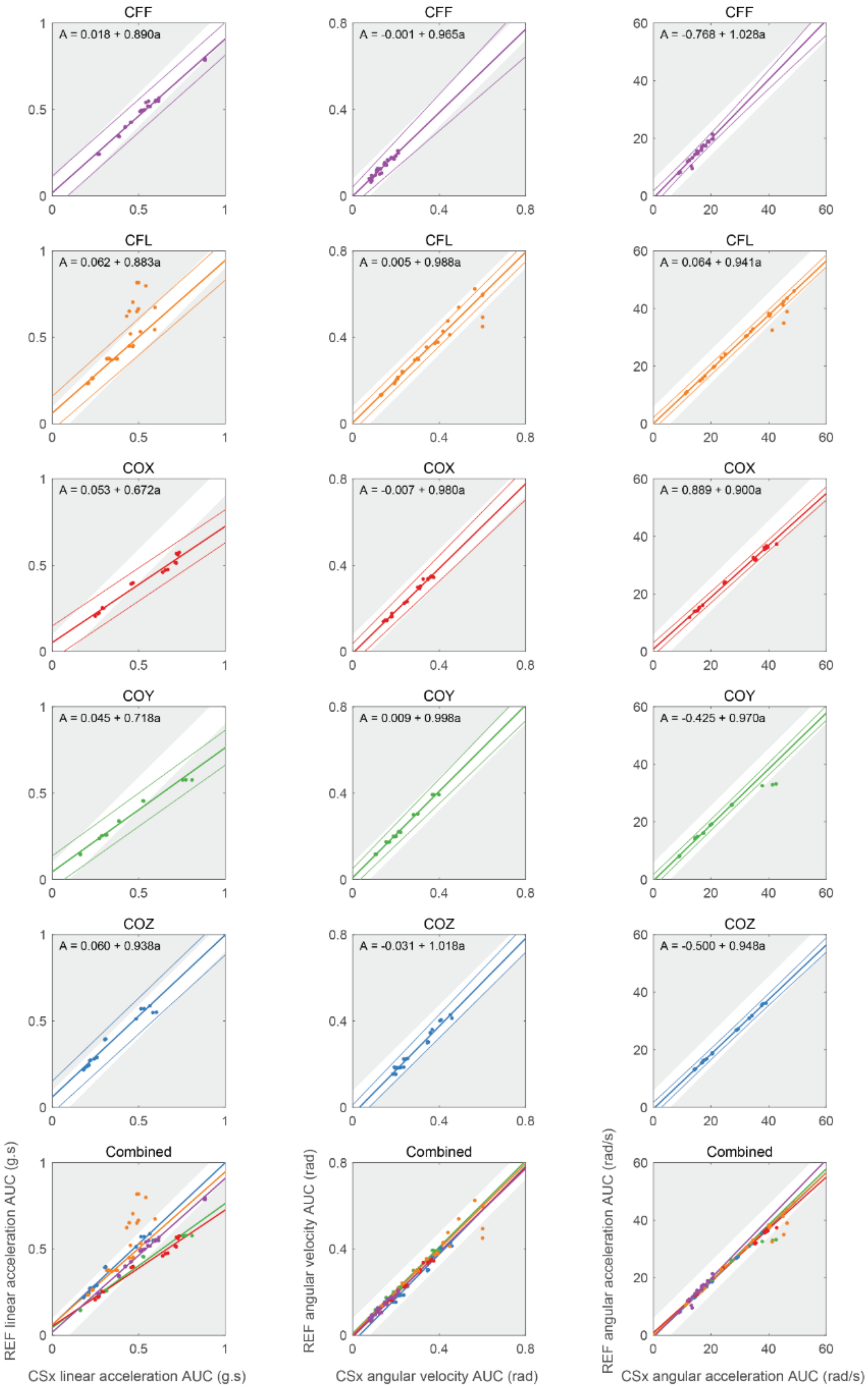


Figure B.15 Linear regression plots of the AUC metric for the linear acceleration, angular velocity and angular acceleration from the 6DOF impacts.

Table B.10 Results of the linear regression models and error means for the linear acceleration, angular velocity and angular acceleration AUCs from the 6DOF impacts (p-value < 0.001 in all cases). Bold values indicate that the slope or intercept are not significantly different from 1 or 0, respectively, or that MAE% is lower than 10%.

		Intercepts				Slopes				Errors	
	N	Estimate	SE	tStat	p*	Estimate	SE	tStat	p*	MAE%	RMSE%
Linear acceleration AUC											
Flat Frontal	26	0.018	0.026	0.674	**0.502**	0.890	0.048	2.274	0.025	**8%**	9%
Flat lateral	27	0.062	0.038	1.153	**0.252**	0.883	0.082	1.426	**0.157**	15%	20%
Oblique coronal	19	0.053	0.037	0.941	**0.349**	0.672	0.068	4.817	<0.001	26%	28%
Oblique sagittal	18	0.045	0.034	0.782	**0.436**	0.718	0.068	4.131	<0.001	19%	20%
Oblique horizontal	21	0.060	0.034	1.248	**0.215**	0.938	0.077	0.805	**0.423**	14%	15%
Angular velocity AUC											
Flat Frontal	26	-0.0006	0.0116	-0.0487	**0.961**	0.9649	0.0833	0.4210	**0.675**	**8%**	12%
Flat lateral	27	0.0049	0.0141	0.3862	**0.700**	0.9877	0.0861	0.1426	**0.887**	**5%**	9%
Oblique coronal	19	-0.0069	0.0178	-0.3591	**0.720**	0.9803	0.0970	0.2033	**0.839**	**4%**	6%
Oblique sagittal	18	0.0093	0.0161	0.6090	**0.544**	0.9976	0.0953	0.0255	**0.980**	**5%**	6%
Oblique horizontal	21	-0.0315	0.0172	-1.7961	**0.075**	1.0176	0.0933	-0.1885	**0.851**	12%	15%
Angular acceleration AUC											
Flat Frontal	26	-0.7677	0.8530	-0.9000	**0.370**	1.0279	0.0541	-0.5156	**0.607**	**5%**	11%
Flat lateral	27	0.0642	0.9847	0.8448	**0.400**	0.9407	0.0561	1.0580	**0.293**	**8%**	10%
Oblique coronal	19	0.8886	1.0393	1.5937	**0.114**	0.9003	0.0577	1.7288	**0.087**	**7%**	8%
Oblique sagittal	18	-0.4245	0.9976	0.3440	**0.732**	0.9699	0.0583	0.5165	**0.607**	**9%**	12%
Oblique horizontal	21	-0.4995	1.0462	0.2563	**0.798**	0.9482	0.0588	0.8802	**0.381**	**8%**	8%

AUC = Area under the curve; SE = standard error; MAE% = Mean absolute error; RMSE% = Root mean square error.
*the p-value for the intercept terms indicates whether it is significantly different from 0, while the p-value for the slope terms indicate whether it is significantly different from 1.

Appendix C - Ethics approval for the sparring study

This appendix reports the ethics approval letter and the participants information sheets and consent form exemplars.

We had defined three types of participants:

- "The boxers": the primary participants of this research from whom head impact data was collected and analysed. They were adult experienced boxers from the Auckland region that regularly took part in sparring. An initial data collection point allowed collecting information about the boxers' training habits and experience, taking measurements of their head and neck and moulding the instrumented mouthguard. The boxers then took part in sparring as part of their normal training.
- "The sparring partners": the secondary participants of this research, who sparred against the boxers. Their actions were analysed from the video footages. They were also adult experienced boxers from the Auckland region that regularly took part in sparring. No data other than video was collected from the sparring partners.
- "The trainers": incidental participants, as the people who supervise the sparring session and may be filmed while being on the side of the ring. No data was analysed from the trainers.

TE WĀNANGA ARONUI O TĀMAKI MAKAU RAU

Auckland University of Technology Ethics Committee (AUTEC)

Auckland University of Technology
D-88, Private Bag 92006, Auckland 1142, NZ
T: +64 9 921 9999 ext. 8316
E: ethics@aut.ac.nz
www.aut.ac.nz/researchethics

5 August 2020

Robert Borotkanics
Faculty of Health and Environmental Sciences

Dear Robert

Re Ethics Application: **20/153 Head impacts monitored using mouth guards, patches and headgear: differences and similarities during sparring.**

Thank you for providing evidence as requested, which satisfies the points raised by the Auckland University of Technology Ethics Committee (AUTEC).

Your ethics application has been approved for three years until 5 August 2023.

Standard Conditions of Approval

1. The research is to be undertaken in accordance with the Auckland University of Technology Code of Conduct for Research and as approved by AUTEC in this application.
2. A progress report is due annually on the anniversary of the approval date, using the EA2 form.
3. A final report is due at the expiration of the approval period, or, upon completion of project, using the EA3 form.
4. Any amendments to the project must be approved by AUTEC prior to being implemented. Amendments can be requested using the EA2 form.
5. Any serious or unexpected adverse events must be reported to AUTEC Secretariat as a matter of priority.
6. Any unforeseen events that might affect continued ethical acceptability of the project should also be reported to the AUTEC Secretariat as a matter of priority.
7. It is your responsibility to ensure that the spelling and grammar of documents being provided to participants or external organisations is of a high standard and that all the dates on the documents are updated.

AUTEC grants ethical approval only. You are responsible for obtaining management approval for access for your research from any institution or organisation at which your research is being conducted and you need to meet all ethical, legal, public health, and locality obligations or requirements for the jurisdictions in which the research is being undertaken.

Please quote the application number and title on all future correspondence related to this project.

For any enquiries please contact ethics@aut.ac.nz. The forms mentioned above are available online through http://www.aut.ac.nz/research/researchethics

(This is a computer-generated letter for which no signature is required)

The AUTEC Secretariat
Auckland University of Technology Ethics Committee

Cc: enora.leflao@aut.ac.nz; Nigel Harris; lenetsky@gmail.com

Participant Information Sheet - Boxers

Date Information Sheet Produced: 01/07/2020

Project Title

Head impacts monitored using mouthguards, patches and headgear: differences and similarities during sparring.

An Invitation

My name is Enora Le Flao and I am a PhD student at Auckland University of Technology (AUT) Sports Performance Research Institute New Zealand (SPRINZ). Along with Dr Robert Borotkanics, my PhD supervisor, we invite you to take part in research investigating head impacts in sparring. I am particularly interested in the technology that allows us to measure head motion after a hit, and how this technology could help us monitor head impacts during training and fights.

This Participant Information Sheet will help you decide if you would like to participate. It sets out why we are doing the research, what your participation would involve, what the benefits and risks to you might be, and what would happen after the research ends.

If you agree to take part in this research, you will be asked to sign a Consent Form. You will be given a copy of both the Participant Information Sheet and the Consent Form to keep.

What is the purpose of this research?

We think that the amount of motion that the head goes through after being hit can affect how you feel and how you perform. We are using head impact sensors (accelerometers, like you have in your phone) to measure this motion during sport. The purpose of this research is to test different sensors and to describe head impacts sustained during sparring.

This research will be undertaken by myself, Enora Le Flao, and will contribute to the fulfilment of my PhD thesis. The results will be published in academic journals and presented at scientific conferences to inform the wider community.

How was I identified and why am I being invited to participate in this research?

You were identified as a potential participant because you are an experienced boxer currently training with a coach we contacted. We thank you for considering being a participant.

How do I agree to participate in this research?

Your participation in this research is voluntary. If you don't want to participate, you don't have to give a reason. If you want to participate now but change your mind later, you can withdraw from the research at any time. Whether or not you choose to participate will neither advantage nor disadvantage you. If you choose to withdraw from the research, then you will be offered the choice between having any data that is identifiable as belonging to you removed or allowing it to continue to be used. However, once the findings have been produced, removal of your data may not be possible.

You do not have to decide today whether you wish to participate in this research. I will be present during your sparring sessions for several weeks. Before you decide you may want to talk about the research with other people, such as family, whānau, friends, or healthcare providers. Feel free to do this and please don't hesitate to ask me any question you may have in relationship to this research.

What will happen in this research?

I will be present during your gym's organised sparring sessions. If you agree to participate in this research, on the first day I will take a few measurements of your head and neck and ask your age, height and weight. You will then answer a few questions about your boxing experience, your training routine and your concussion history. Finally, we will mould an instrumented mouthguard on your teeth (see figure below), just as you would do with a regular boil-and-bite mouthguard.

The next time we meet at one of your sparring sessions, I will give you your mouthguard and equip you with two other head impact sensors. One I will tape behind your right ear (see figure below), and the second on your headgear. I will take a few pictures of your head and face with the sensors. You will then participate in the sparring session as part of your normal training routine. The sensor will record head motion throughout the session. This will also be videoed using three cameras located around the ring, that I will watch and analyse later on. This is to ensure that what is recorded by the sensors corresponds to punches that I can see on video. After a sparring session, I will also ask you a few questions about the intensity of the session and how you are feeling.

These sparring sessions with the sensors can be repeated several times if you agree to participate (ideally three times). It should only take a few minutes before and after you are in the ring for me to attach and remove the sensors.


Examples of the sensors used in the research: Left: an unmoulded instrumented mouthguard; Right: a patch taped behind the right ear (the same patch will be taped on the headgear worn by the boxer). The dimensions of the patches are 26x21x7 mm.

What are the discomforts and risks?

The main risks that can occur are those that normally occur from participating in sparring. This includes the risk of concussion, and this risk is increased if you have had a previous concussion. If you feel symptoms of concussion after sparring or fighting (headache, feeling like in a fog, amnesia, balance issues, sleep/wake disturbance, irritability), it is recommended you consult your GP. Rest for at least 24 hours, then return to sport gradually: increase the intensity of the exercise only if you are symptom-free. We invite you to consult ACC's guidelines on concussion management on https://www.accsportsmart.co.nz/concussion. These guidelines are summarised by three items: (1) Recognise the symptoms; (2) Remove from play; (3) Refer to a doctor for assessment.

With respect to the sensors, they all have been utilised in sports before with no reported safety issue. Some discomfort may be caused by the mouthguard being slightly bulkier than your usual mouthguard (this is due to the electronics on board), or by the patch being taped on your skin behind your ear. I will use a strong wig tape that should resist to sweat and may have to shave a small square of hair to make the sensor stick better and to facilitate removal.

How will these discomforts and risks be alleviated?

We will ask you to wear competition-approved headgear for this study.

If you are uncomfortable with me, a woman, taking your head measurements and fixing the sensor behind your ear, please let me know and I will ensure a male researcher will be there with me.

If you are feeling distressed as a result of this research, AUT Health Counselling and Wellbeing is able to offer three free sessions of confidential counselling support. These sessions are only available for issues that have arisen directly as a result of participation in the research and are not for other general counselling needs. To access these services, you will need to:

- drop into our centres at WB219 or AS104 or phone 921 9992 City Campus or 921 9998 North Shore campus to make an appointment. Appointments for South Campus can be made by calling 921 9992
- let the receptionist know that you are a research participant, and provide the title of my research and my name and contact details as given in this Information Sheet

You can find out more information about AUT counsellors and counselling on http://www.aut.ac.nz/being-a-student/current-postgraduates/your-health-and-wellbeing/counselling.

What are the benefits?

Information gained from this research will help understand the differences between the sensors and may indirectly help in their improvement. We will also gain a better understanding of head motion during sparring and relative to the different punches you received. Ultimately, we may be able to use this knowledge and those sensors to develop tools to monitor a boxer's load in terms of head contacts sustained during training and fighting (number, type, and magnitude of head contacts). Finally, this research will help me personally obtain a PhD degree.

What compensation is available for injury or negligence?

In the event of a physical injury as a result of your participation in this study, rehabilitation and compensation for injury by accident may be available from the Accident Compensation Corporation, providing the incident details satisfy the requirements of the law and the Corporation's regulations.

How will the data be used?

The video footages will be used to observe landed punches. We will use this information, as well as your personal information (boxing experience, neck size, etc.) to analyse the data from the head impact sensors.

I may share the data with other researchers to be used for more detailed analysis than what I can do within the scope of my PhD. This would consist in characterising the punches landed during the session (e.g. punch type, direction, where it landed). You may choose to consent or not to your data being shared by signing the consent form.

I will modify any image and video that may be used for scientific communication to ensure that you are not identifiable. You may choose to consent or not to your de-identified image being used outside of this specific study by signing the consent form.

If you wish and if your sparring partner and the session's referee agree, you will be able to access the videos of your session. Similarly, you will be able to decide if you wish your video to be shared with them.

How will my privacy be protected?

The research involves participation at your gym where other people will be training. We will video sessions. Therefore, we are unable to guarantee your privacy during data collection. You are free to choose what you allow us to do with your data by signing the consent form.

Your sparring partner may ask for a copy of the video for training purposes, which you can consent to or not by signing the consent form.

What are the costs of participating in this research?

Participating in this research project will not cost you apart from the time that you will spend to answer the questions, get your measurements taken and be equipped with the sensor. The research has been designed to minimally disrupt your usual training routine.

What opportunity do I have to consider this invitation?

Please take the necessary time you need to consider the invitation to participate in this research. I will be present at your gym several times over the next few weeks.

Will I receive feedback on the results of this research?

You can choose to receive a 1-page summary of the results at the end of the research, by ticking the appropriate box on the Consent Form accompanying this Information Sheet.

What do I do if I have concerns about this research?

Any concerns regarding the nature of this project should be notified in the first instance to the Project Supervisor, Dr Robert Borotkanics. Email: robert.borotkanics@aut.ac.nz .

Concerns regarding the conduct of the research should be notified to the Executive Secretary of AUTEC, *ethics@aut.ac.nz* , (+649) 921 9999 ext 6038.

Whom do I contact for further information about this research?

Please keep this Information Sheet and a copy of the Consent Form for your future reference. You are also able to contact the research team as follows:

Researcher Contact Details:

Enora Le Flao, Sports Performance Research Institute New Zealand, School of Sport and Recreation, Auckland University of Technology. Email: enora.leflao@aut.ac.nz .

Enora Le Flao has received no external funding for this research nor has been influenced to undertake this research by third parties.

Project Supervisor Contact Details:

Dr Robert Borotkanics, Sports Performance Research Institute New Zealand, School of Sport and Recreation, Auckland University of Technology. Email: robert.borotkanics@aut.ac.nz .

Approved by the Auckland University of Technology Ethics Committee on 5 August 2020, AUTEC Reference number 20/153.

TE WĀNANGA ARONUI O TĀMAKI MAKAU RAU

Consent and Release Form - Boxer

Project title: ***Head impacts monitored using mouthguards, patches and headguard: differences and similarities during sparring.***

Project Supervisor: ***Dr Robert Borotkanics***

Researcher: ***Enora Le Flao***

- ☐ I have read and understood the information provided about this research project in the Information Sheet dated 1st July 2020.
- ☐ I have had an opportunity to ask questions and to have them answered.
- ☐ I understand that taking part in this study is voluntary (my choice) and that I may withdraw from the study at any time without being disadvantaged in any way.
- ☐ I understand that if I withdraw from the study then I will be offered the choice between having any data or tissue that is identifiable as belonging to me removed or allowing it to continue to be used. However, once the findings have been produced, removal of my data may not be possible.
- ☐ I authorize the researcher to video or otherwise make a video reproduction of me and/or record my voice.
- ☐ I understand that my privacy cannot be guaranteed during data collection, as data will be collected at gyms where other people may be training.
- ☐ I understand that the photos and videos will be used for academic purposes and will not be published in any form outside of this project without my written permission.
- ☐ I permit the researcher to use the de-identified photos and videos, either complete or in part, alone or in conjunction with any wording and/or drawings solely and exclusively for (a) the researcher's portfolio; (b) dissemination of research findings to the scientific community.
- ☐ I am free of any injury that would prevent me from participating in sparring and have not sustained a head injury in the last year.
- ☐ I agree to take part in this research.

I agree to the data being shared with other researchers for more detailed analysis (analysis of the videos to characterise punches).	Yes ☐	No ☐
I permit the researcher to give a copy of the videos to my sparring partner.	Yes ☐	No ☐
I permit the researcher to give a copy of the videos to my trainer.	Yes ☐	No ☐
I wish to receive a summary of the research findings.	Yes ☐	No ☐

Participant's signature: ....

Participant's name: ....

Participant's Contact Details (if appropriate):

Email: ....

Date: ....

Approved by the Auckland University of Technology Ethics Committee on 5 August 2020, AUTEC Reference number 20/153.

Note: The Participant should retain a copy of this form.

April 2018 page 1 of 1 This version was last edited in April 2018

TE WĀNANGA ARONUI O TĀMAKI MAKAU RAU

Participant Information Sheet – Sparring partner

Date Information Sheet Produced: 01/07/2020

Project Title

Head impacts monitored using mouthguards, patches and headguard: differences and similarities during sparring.

An Invitation

My name is Enora Le Flao and I am a PhD student at Auckland University of Technology (AUT) Sports Performance Research Institute New Zealand (SPRINZ). Along with Dr Robert Borotkanics, my PhD supervisor, we invite you to take part in research investigating head impacts in sparring. I am particularly interested in the technology that allows us to measure head motion after a hit, and how this technology could help us monitor head impacts during training and fights.

This Participant Information Sheet will help you decide if you would like to participate. It sets out why we are doing the research, what your participation would involve, what the benefits and risks to you might be, and what would happen after the research ends.

If you agree to take part in this research, you will be asked to sign a Consent Form. You will be given a copy of both the Participant Information Sheet and the Consent Form to keep.

What is the purpose of this research?

We think that the amount of motion that the head goes through after being hit can affect how you feel and you perform. We are using head impact sensors (accelerometers, like you have in your phone) to measure this motion during sport. The purpose of this research is to test different sensors and to describe head impacts sustained in the ring.

This research will be undertaken by myself, Enora Le Flao, and will contribute to the fulfilment of my PhD thesis. The results will be published in academic journals and presented at scientific conferences to inform the wider community.

How was I identified and why am I being invited to participate in this research?

You were identified as a potential participant because you are sparring with one of our study participants.

How do I agree to participate in this research?

Your participation in this research is voluntary. If you don't want to participate, you don't have to give a reason. If you want to participate now but change your mind later, you can withdraw from the research at any time. Whether or not you choose to participate will neither advantage nor disadvantage you. If you choose to withdraw from the research, then you will be offered the choice between having any data that is identifiable as belonging to you removed or allowing it to continue to be used. However, once the findings have been produced, removal of your data may not be possible.

You do not have to decide today whether you wish to participate in this research. Before you decide you may want to talk about the research with other people, such as family, whānau, friends, or healthcare providers. Feel free to do this. Please don't hesitate to ask me any question you may have in relationship to this research.

What will happen in this research?

I am collecting information during the sparring session(s) using sensors placed on the other boxer's head. I am also filming these sessions using several cameras covering the ring in order to analyse what punches that boxer sustained. This means that you will be filmed. Later on, I will analyse the videos of the sparring session to identify what type of punches you thrown and landed on your opponents.

What are the discomforts and risks?

You will participate in sparring as you would normally do, therefore there is no increased risks because of the study. You may feel uncomfortable by the idea of having someone (me, the primary researcher) analysing videos where you can be identified.

There are various injury risks from participating in sparring. This includes the risk of concussion, and this risk is increased if you have had a previous concussion. If you feel symptoms of concussion after sparring or fighting (headache, feeling like in a fog, amnesia, balance issues, sleep/wake disturbance, irritability), it is recommended you consult your GP. Rest for at least 24 hours, then return to sport gradually: increase the intensity of your exercise only if you are symptom-free. We invite you to consult ACC's guidelines on concussion management on https://www.accsportsmart.co.nz/concussion. These guidelines are summarised by three items: (1) Recognise the symptoms; (2) Remove from play; (3) Refer to a doctor for assessment.

How will these discomforts and risks be alleviated?

We will ask you to wear competition-approved headgear for this study.

If you are feeling distressed as a result of this research, AUT Health Counselling and Wellbeing is able to offer three free sessions of confidential counselling support. These sessions are only available for issues that have arisen directly as a result of participation in the research and are not for other general counselling needs. To access these services, you will need to:

- drop into our centres at WB219 or AS104 or phone 921 9992 City Campus or 921 9998 North Shore campus to make an appointment. Appointments for South Campus can be made by calling 921 9992
- let the receptionist know that you are a research participant, and provide the title of my research and my name and contact details as given in this Information Sheet

You can find out more information about AUT counsellors and counselling on http://www.aut.ac.nz/being-a-student/current-postgraduates/your-health-and-wellbeing/counselling.

What are the benefits?

Information gained from this research will help understand the differences between the sensors and may indirectly help in their improvement. We will also gain a better understanding of head motion during sparring and relative to the different punches you received. Ultimately, we may be able to use this knowledge and those sensors to develop tools to monitor a boxer's load in terms of head contacts sustained during training and fighting (number, type, and magnitude of head contacts). Finally, this research will help me personally obtain a PhD degree.

What compensation is available for injury or negligence?

In the event of a physical injury as a result of your participation in this study, rehabilitation and compensation for injury by accident may be available from the Accident Compensation Corporation, providing the incident details satisfy the requirements of the law and the Corporation's regulations.

How will the data be used?

The video footages will be used to observe landed punches and their characteristics (e.g. punch type, direction, where it landed). We will use this information to analyse the data from the head impact sensors.

I may share the data with other researchers to be used for more detailed analysis than what I can do within the scope of my PhD. You may choose to consent or not to your data being shared by signing the consent form.

I will modify any image and video that may be used for scientific communication to ensure that you are not identifiable. You may choose to consent or not to your de-identified image being used outside of this specific study by signing the consent form.

If you wish and if your sparring partner and the session's referee agree, you will be able to access the videos of your session. Similarly, you will be able to decide if you wish your video to be shared with them.

How will my privacy be protected?

The research involves participation at your gym where other people will be training. We will video sessions. Therefore, we are unable to guarantee your privacy during data collection. However, you are free to choose what you allow us to do with your data by signing the consent form.

Your sparring partner may ask for a copy of the video for training purposes, which you can consent to or not by signing the consent form. I you wish and if your sparring partner agrees, you will be able to access the videos of your session.

What are the costs of participating in this research?

There are no costs to you of participating in this research. The research has been designed to minimally disrupt your usual training routine.

What opportunity do I have to consider this invitation?

You should have received this information sheet ahead of the sparring session we are recording, to allow you time for reflection. Whether you choose to participate in this study or not, I will need your answer before the sparring session with the other boxers I am working with starts.

Will I receive feedback on the results of this research?

You can choose to receive a 1-page summary of the results at the end of the research, by ticking the appropriate box on the Consent Form accompanying this Information Sheet.

What do I do if I have concerns about this research?

Any concerns regarding the nature of this project should be notified in the first instance to the Project Supervisor, Dr Robert Borotkanics. Email: robert.borotkanics@aut.ac.nz .

Concerns regarding the conduct of the research should be notified to the Executive Secretary of AUTEC, *ethics@aut.ac.nz* , (+649) 921 9999 ext 6038.

Whom do I contact for further information about this research?

Please keep this Information Sheet and a copy of the Consent Form for your future reference. You are also able to contact the research team as follows:

Researcher Contact Details:

Enora Le Flao, Sports Performance Research Institute New Zealand, School of Sport and Recreation, Auckland University of Technology. Email: enora.leflao@aut.ac.nz .

Enora Le Flao has received no external funding for this research nor has been influenced to undertake this research by third parties.

Project Supervisor Contact Details:

Dr Robert Borotkanics, Sports Performance Research Institute New Zealand, School of Sport and Recreation, Auckland University of Technology. Email: robert.borotkanics@aut.ac.nz .

Approved by the Auckland University of Technology Ethics Committee on 5 August 2020, AUTEC Reference number 20/153.

TE WĀNANGA ARONUI O TĀMAKI MAKAU RAU

Consent and Release Form – Sparring partner

Project title: ***Head impacts monitored using mouthguards, patches and headguard: differences and similarities during sparring.***

Project Supervisor: ***Dr Robert Borotkanics***

Researcher: ***Enora Le Flao***

- ☐ I have read and understood the information provided about this research project in the Information Sheet dated 1st July 2020.
- ☐ I have had an opportunity to ask questions and to have them answered.
- ☐ I authorize the researcher to video or otherwise make a video reproduction of me and/or record my voice.
- ☐ I understand that being videoed for this study is voluntary (my choice) and that I may withdraw my consent at any time without being disadvantaged in any way.
- ☐ I understand that if I withdraw my consent then I will be offered the choice between having any data that is identifiable as belonging to me removed or allowing it to continue to be used. However, once the findings have been produced, removal of my data may not be possible.
- ☐ I understand that the photos and videos will be used for academic purposes and may be published outside of this project.
- ☐ I permit the researcher to use the de-identified photos and videos, either complete or in part, alone or in conjunction with any wording and/or drawings solely and exclusively for (a) the researcher's portfolio; (b) dissemination of research findings to the scientific community.
- ☐ I am free of any injury that would prevent me from participating in sparring and have not sustained a head injury in the last year.
- ☐ I agree to take part in this research.

I agree to the data being shared with other researchers for more detailed analysis (analysis of the videos to characterise punches).	Yes ☐	No ☐
I permit the researcher to give a copy of the videos to my sparring partner.	Yes ☐	No ☐
I permit the researcher to give a copy of the videos to my trainer.	Yes ☐	No ☐
I wish to receive a summary of the research findings.	Yes ☐	No ☐

Participant's signature: ....

Participant's name: ....

Participant's Contact Details (if appropriate):

Email: ....

Date: ....

Approved by the Auckland University of Technology Ethics Committee on 5 August 2020, AUTEC Reference number 20/153.

Note: The Participant should retain a copy of this form.

April 2018 page 1 of 1 This version was last edited in April 2018

TE WĀNANGA ARONUI O TĀMAKI MAKAU RAU

Participant Information Sheet – Trainer

Date Information Sheet Produced: 01/07/2020

Project Title

Head impacts monitored using mouthguards, patches and headguard: differences and similarities during sparring.

An Invitation

My name is Enora Le Flao and I am a PhD student at Auckland University of Technology (AUT) Sports Performance Research Institute New Zealand (SPRINZ). Along with Dr Robert Borotkanics, my PhD supervisor, we invite you to take part in research investigating head impacts in sparring. I am particularly interested in the technology that allows us to measure head motion after a hit, and how this technology could help us monitor head impacts during training and fights.

This Participant Information Sheet will help you decide if you would like to participate. It sets out why we are doing the research, what your participation would involve, what the benefits and risks to you might be, and what would happen after the research ends.

If you agree to take part in this research, you will be asked to sign a Consent Form. You will be given a copy of both the Participant Information Sheet and the Consent Form to keep.

What is the purpose of this research?

We think that the amount of motion that the head goes through after being hit can affect how you feel and you perform. We are using head impact sensors (accelerometers, like you have in your phone) to measure this motion during sport. The purpose of this research is to test different sensors and to describe head impacts sustained in the ring.

This research will be undertaken by myself, Enora Le Flao, and will contribute to the fulfilment of my PhD thesis. The results will be published in academic journals and presented at scientific conferences to inform the wider community.

How was I identified and why am I being invited to participate in this research?

You were identified as a potential participant because you are supervising a sparring session that involves our study participants.

How do I agree to participate in this research?

Your participation in this research is voluntary. If you don't want to participate, you don't have to give a reason. If you want to participate now but change your mind later, you can withdraw from the research at any time. Whether or not you choose to participate will neither advantage nor disadvantage you. If you choose to withdraw from the research, then you will be offered the choice between having any data that is identifiable as belonging to you removed or allowing it to continue to be used. However, once the findings have been produced, removal of your data may not be possible.

You do not have to decide today whether you wish to participate in this research. Before you decide you may want to talk about the research with other people, such as family, whānau, friends, or healthcare providers. Feel free to do this. Please don't hesitate to ask me any question you may have in relationship to this research.

What will happen in this research?

I am collecting information during the sparring session(s) using sensors and several cameras covering the ring in order to analyse what punches the boxers sustained. This means that you will be filmed as well, but there will be no analysis of your actions.

What are the discomforts and risks?

You will participate in supervising/watching the bout as you would normally do, therefore there is no increased risks or discomfort because of the study.

What are the benefits?

Information gained from this research will help understand the differences between the sensors and may indirectly help in their improvement. We will also gain a better understanding of head motion during sparring and relative to the different punches you received. Ultimately, we may be able to use this knowledge and those sensors to develop tools to monitor a boxer's load in terms of head contacts sustained during training and fighting (number, type, and magnitude of head contacts). Finally, this research will help me personally obtain a PhD degree.

What compensation is available for injury or negligence?

In the event of a physical injury as a result of your participation in this study, rehabilitation and compensation for injury by accident may be available from the Accident Compensation Corporation, providing the incident details satisfy the requirements of the law and the Corporation's regulations.

How will the data be used?

The video footages will be used to observe landed punches and their characteristics (e.g. punch type, direction, where it landed). We will use this information to analyse the data from the head impact sensors.

I may share the data with other researchers to be used for more detailed analysis than what I can do within the scope of my PhD. You may choose to consent or not to your video being shared by signing the consent form.

I will modify any image and video that may be used for scientific communication to ensure that you are not identifiable. You may choose to consent or not to your de-identified image being used outside of this specific study by signing the consent form.

If you wish and if the boxers agree, you will be able to access the videos of the session. Similarly, you will be able to decide if you wish your video to be shared with them.

How will my privacy be protected?

The research involves participation at your gym where other people will be training. We will video sessions. Therefore, we are unable to guarantee your privacy during data collection. However, you are free to choose what you allow us to do with your data by signing the consent form.

The boxers may ask for a copy of the video for training purposes, which you can consent to or not by signing the consent form.

What are the costs of participating in this research?

There are no costs to you of participating in this research. The research has been designed to minimally disrupt your usual training routine.

What opportunity do I have to consider this invitation?

Whether you choose to participate in this study or not, I will need your answer before the sparring session with the other boxers I am working with starts.

Will I receive feedback on the results of this research?

You can choose to receive a 1-page summary of the results at the end of the research, by ticking the appropriate box on the Consent Form accompanying this Information Sheet.

What do I do if I have concerns about this research?

Any concerns regarding the nature of this project should be notified in the first instance to the Project Supervisor, Dr Robert Borotkanics. Email: robert.borotkanics@aut.ac.nz .

Concerns regarding the conduct of the research should be notified to the Executive Secretary of AUTEC, *ethics@aut.ac.nz* , (+649) 921 9999 ext 6038.

Whom do I contact for further information about this research?

Please keep this Information Sheet and a copy of the Consent Form for your future reference. You are also able to contact the research team as follows:

Researcher Contact Details:

Enora Le Flao, Sports Performance Research Institute New Zealand, School of Sport and Recreation, Auckland University of Technology. Email: enora.leflao@aut.ac.nz .

Enora Le Flao has received no external funding for this research nor has been influenced to undertake this research by third parties.

Project Supervisor Contact Details:

Dr Robert Borotkanics, Sports Performance Research Institute New Zealand, School of Sport and Recreation, Auckland University of Technology. Email: robert.borotkanics@aut.ac.nz .

Approved by the Auckland University of Technology Ethics Committee on 5 August 2020, AUTEC Reference number 20/153.

TE WĀNANGA ARONUI O TĀMAKI MAKAU RAU

Consent and Release Form – Trainer

Project title:	***Head impacts monitored using mouthguards, patches and headguard: differences and similarities during sparring.***
Project Supervisor:	***Dr Robert Borotkanics***
Researcher:	***Enora Le Flao***

- ☐ I have read and understood the information provided about this research project in the Information Sheet dated 1st July 2020.
- ☐ I have had an opportunity to ask questions and to have them answered.
- ☐ I authorize the researcher to video or otherwise make a video reproduction of me and/or record my voice.
- ☐ I understand that being videoed for this study is voluntary (my choice) and that I may withdraw my consent at any time without being disadvantaged in any way.
- ☐ I understand that if I withdraw my consent then I will be offered the choice between having any data that is identifiable as belonging to me removed or allowing it to continue to be used. However, once the findings have been produced, removal of my data may not be possible.
- ☐ I understand that the photos and videos will be used for academic purposes and may be published outside of this project.
- ☐ I permit the researcher to use the de-identified photos and videos, either complete or in part, alone or in conjunction with any wording and/or drawings solely and exclusively for (a) the researcher's portfolio; (b) dissemination of research findings to the scientific community.
- ☐ I understand that my athletes have a choice to participate or not in this study. Their decision will not be influenced by me and will be respected.

I agree to the data being shared with other researchers for more detailed analysis (analysis of the videos to characterise punches).	Yes ☐	No ☐
I permit the researcher to give a copy of the videos to the boxers.	Yes ☐	No ☐
I wish to receive a summary of the research findings.	Yes ☐	No ☐

Participant's signature: ....

Participant's name: ....

Participant's Contact Details (if appropriate):

Email: ....

Date: ....

Approved by the Auckland University of Technology Ethics Committee on 5 August 2020, AUTEC Reference number 20/153.

Note: The Participant should retain a copy of this form.

April 2018 page 1 of 1 This version was last edited in April 2018

Appendix D - Elements of protocol and accompanying documents for the sparring study

This appendix includes the links to the various questionnaires given to the participants and the questionnaire itself for the short concussion symptoms assessments, the logs used during the data collection session, the Prevent Biometrics mouthguard fitting instructions and the end-of-study questionnaire about the comfort of the sensors.

Participant information questionnaire

The participants information, training history and habits and concussion history were recorded via an online survey that can be accessed at the address below or using the QR code. The structure of the survey is described in Table D.1.

https://aut.au1.qualtrics.com/jfe/form/SV_e9HCU5jR4L5bflb

Table D.1. Description of the structure of the participants questionnaire with key references used to design the questionnaire.

Survey blocks	Relevance	Key references
Information sheet and consent	Explain the purpose of the survey, obtain consent	
Demographics: sex, age, locality	Describe the population	
Sports loop (repeated for each combat sports the participant has participated in): experience, records and KO: for all the full-contact combat sports practiced over the years: years of experience, competition activity and records, times they have been knocked out with loss of consciousness, and times they have been declared unfit to continue sparring/fighting.	Estimate the overall exposure to combat sports and concussions as it would be defined by competition officials.	
Diagnosed concussions: for all the times they have been KO or declared unfit to continue, have they sought medical attention and were concussions diagnosed. Other concussions in their life, treatment and rehabilitation.	Estimate the lifetime number of diagnosed concussions. Assess the dichotomy between sustained concussion and concussion symptoms. Assess if a history of concussions increases the number/frequency of symptoms.	85
Definition of concussion and symptoms.	Define what a concussion is according to the latest knowledge, informing them along the way.	6, 341
Number of concussions: recall of concussions over their lifetime, while competing or sparring. Frequency of the duration of their recovery, and duration before they went back to full-contact sparring.	Estimate a more realistic number of concussions and estimate the severity of it.	30
Sparring & Symptoms loop (repeated for the sports that have been practiced regularly in the last 2 years, to prevent recall bias), training and sparring habits: frequency, duration, protective equipment, intensity, intensity of the strikes to the head. Frequency of symptoms, duration of symptoms and consequences.	Evaluate the exposure in terms of training load and intensity. Assess how participation in sparring affects personal life, ability to work/study, performance on the ring.	126, 159, 185, 256
Attitude and behaviour regarding concussion: pushing through symptoms, non-disclosure of symptoms, following medical advice if applicable, severity of symptoms, awareness of concussion-related risks and protocols.	Understanding the attitudes of athletes and identifying education pathways.	30, 74, 126, 159
Other risk factors: history of other potential risk factors for concussion, neck strengthening routine.	Assess if these factors influence the frequency or severity of concussion-related symptoms.	
Thanks and additional information: link to a blank webpage where results will be available later, links to concussion-related resources and guidelines.		286, 291

Data collection day log

☐ Prevent Biometrics teambox plugged, all sensors charging

☐ Prevent Biometrics app open and access to the Internet via Wifi

☐ Blue-tac on board ready

☐ CSx team box ON, GPS time set, all sensors OK and charging

☐ CSx team box, iPad and watch all showing similar time: ....

☐ Cameras ready

	PB mouthguards								CSx patch SKIN					CSx patch HEADGEAR				
	03	04	05	06	07	08	09	10	01	02	03	04	05	06	07	08	09	10
Sensors out of box and no sign of damage																		
Receiving impacts																		
Charge status																		
Blue-tac on and sensors to board																		
☐ Master camera ON at																		
Filmed the change of minutes	☐ iPad time .... Video								☐ Team box time .... Video									
☐ Calibration impact START on iPad .... On video																		
Mouthguards to Trigene																		
Sensors on/with participant																		
Photographs and measurements taken																		
Data collection																		
☐ Master camera still ON?																		
Sensors collected, tape taken off, to Trigene																		
Blue-tac on and sensors to board																		
☐ Calibration impact END on iPad .... On video																		
☐ Master camera OFF at .... Video duration																		
All tasks OK / Downloaded impacts																		
Charge status																		

Sparring rounds log

Round	Start and end times	Participant	Participant	Round	Start and end times	Participant	Participant
01		Ring / Out / Rest Filmed / Not SK HG PA	Ring / Out / Rest Filmed / Not SK HG PA	11		Ring / Out / Rest Filmed / Not SK HG PA	Ring / Out / Rest Filmed / Not SK HG PA
02		Ring / Out / Rest Filmed / Not SK HG PA	Ring / Out / Rest Filmed / Not SK HG PA	12		Ring / Out / Rest Filmed / Not SK HG PA	Ring / Out / Rest Filmed / Not SK HG PA
03		Ring / Out / Rest Filmed / Not SK HG PA	Ring / Out / Rest Filmed / Not SK HG PA	13		Ring / Out / Rest Filmed / Not SK HG PA	Ring / Out / Rest Filmed / Not SK HG PA
04		Ring / Out / Rest Filmed / Not SK HG PA	Ring / Out / Rest Filmed / Not SK HG PA	14		Ring / Out / Rest Filmed / Not SK HG PA	Ring / Out / Rest Filmed / Not SK HG PA
05		Ring / Out / Rest Filmed / Not SK HG PA	Ring / Out / Rest Filmed / Not SK HG PA	15		Ring / Out / Rest Filmed / Not SK HG PA	Ring / Out / Rest Filmed / Not SK HG PA
06		Ring / Out / Rest Filmed / Not SK HG PA	Ring / Out / Rest Filmed / Not SK HG PA	16		Ring / Out / Rest Filmed / Not SK HG PA	Ring / Out / Rest Filmed / Not SK HG PA
07		Ring / Out / Rest Filmed / Not SK HG PA	Ring / Out / Rest Filmed / Not SK HG PA	17		Ring / Out / Rest Filmed / Not SK HG PA	Ring / Out / Rest Filmed / Not SK HG PA
08		Ring / Out / Rest Filmed / Not SK HG PA	Ring / Out / Rest Filmed / Not SK HG PA	18		Ring / Out / Rest Filmed / Not SK HG PA	Ring / Out / Rest Filmed / Not SK HG PA
09		Ring / Out / Rest Filmed / Not SK HG PA	Ring / Out / Rest Filmed / Not SK HG PA	19		Ring / Out / Rest Filmed / Not SK HG PA	Ring / Out / Rest Filmed / Not SK HG PA
10		Ring / Out / Rest Filmed / Not SK HG PA	Ring / Out / Rest Filmed / Not SK HG PA	20		Ring / Out / Rest Filmed / Not SK HG PA	Ring / Out / Rest Filmed / Not SK HG PA

Concussion symptoms – Pre-bout questionnaire

https://aut.au1.qualtrics.com/jfe/form/SV_1TaQOTL7QCR5SSx

Start of Block: First page

Q1.1 This short survey will assess how you feel before your session of competitive sparring, as a baseline measurement that I will compare your post-sparing response to.

Please answer honestly. Your answers will be kept private.

Q1.2 Name or participant number*

Start of Block: Symptoms

Q2.1 Do you feel any of the symptoms right now?

Please rate their intensity (0 if you are not experiencing it).

	0 - None	1	2 – Mild	3	4 – Moderate	5	6 - Severe
Headache (1)	○	○	○	○	○	○	○
"Pressure in head" (2)	○	○	○	○	○	○	○
Neck pain (3)	○	○	○	○	○	○	○
Nausea or vomiting (4)	○	○	○	○	○	○	○
Dizziness (5)	○	○	○	○	○	○	○
Blurred vision (6)	○	○	○	○	○	○	○
Balance problems (7)	○	○	○	○	○	○	○
Sensitivity to light (8)	○	○	○	○	○	○	○
Sensitivity to noise (9)	○	○	○	○	○	○	○
Feeling slowed down (10)	○	○	○	○	○	○	○
Feeling like in a fog (11)	○	○	○	○	○	○	○
"Don't feel right" (12)	○	○	○	○	○	○	○
Difficulty concentrating (13)	○	○	○	○	○	○	○
Difficulty remembering (14)	○	○	○	○	○	○	○
Fatigue or low energy (15)	○	○	○	○	○	○	○
Confusion (16)	○	○	○	○	○	○	○
Drowsiness (17)	○	○	○	○	○	○	○
More emotional (18)	○	○	○	○	○	○	○
Irritability (19)	○	○	○	○	○	○	○
Sadness (20)	○	○	○	○	○	○	○
Nervous or anxious (21)	○	○	○	○	○	○	○

Q2.2 If 100% is feeling perfectly normal, what percent of normal do you feel?*

Concussion symptoms – Post-bout (immediate) questionnaire

https://aut.au1.qualtrics.com/jfe/form/SV_4IPeKeq7H9jDAKF

Start of Block: First page

Q1.1 This short survey will assess how you feel after your session of competitive sparring. This will allow me to compare it to the measurements taken with the sensors during the bout.
Please answer honestly. Your answers will be kept private.

Q1.2 Name or participant number*

Start of Block: Bout intensity – will loop 5 times to cover all the bouts of the day

Q2.1 Who was it against, on/off the ring?

Q2.2 Intensity

- o Minimal intensity (1)
- o Moderate intensity (2)
- o Intense, but not quite as intense as competition (3)
- o As intense as competition (4)
- o NA (5)

Q2.3 Intensity of the strikes to the head

- o Extremely light (1)
- o Light (2)
- o Moderate (3)
- o Heavy (4)
- o NA (5)

Q2.4 Have you taken one (or more) heavy blow to the head that rocked or fazed you, or left you feeling dizzy, off-balanced or with blurry vision, even for a short time?

- o Yes (1)
- o No (2)

Q2.5 Do you remember which round it was? The type of punch? (conditional on Q2.4)

Start of Block: Video access

Q3.1 Would you like to receive some of the footage from today's session?

Start of Block: Symptoms

Q4.1 Do you feel any of the symptoms below right now?

Please rate their intensity (0 if you are not experiencing it).

	0 - None	1	2 – Mild	3	4 – Moderate	5	6 - Severe
Headache (1)	○	○	○	○	○	○	○
"Pressure in head" (2)	○	○	○	○	○	○	○
Neck pain (3)	○	○	○	○	○	○	○
Nausea or vomiting (4)	○	○	○	○	○	○	○
Dizziness (5)	○	○	○	○	○	○	○
Blurred vision (6)	○	○	○	○	○	○	○
Balance problems (7)	○	○	○	○	○	○	○
Sensitivity to light (8)	○	○	○	○	○	○	○
Sensitivity to noise (9)	○	○	○	○	○	○	○
Feeling slowed down (10)	○	○	○	○	○	○	○
Feeling like in a fog (11)	○	○	○	○	○	○	○
"Don't feel right" (12)	○	○	○	○	○	○	○
Difficulty concentrating (13)	○	○	○	○	○	○	○
Difficulty remembering (14)	○	○	○	○	○	○	○
Fatigue or low energy (15)	○	○	○	○	○	○	○
Confusion (16)	○	○	○	○	○	○	○
Drowsiness (17)	○	○	○	○	○	○	○
More emotional (18)	○	○	○	○	○	○	○
Irritability (19)	○	○	○	○	○	○	○
Sadness (20)	○	○	○	○	○	○	○
Nervous or anxious (21)	○	○	○	○	○	○	○

Q4.2 If 100% is feeling perfectly normal, what percent of normal do you feel?*

Concussion symptoms – Post-bout (48hrs) questionnaire

https://aut.au1.qualtrics.com/jfe/form/SV_0Bc2cpCOIonnd4h

Start of Block: First page

Q1.1 This short survey will assess how you have felt in the 48 hours following your session of competitive sparring. The first section will focus on the moment you have felt the worst, and the second section on approximately 48 hours after the session (~Monday mid-day).

This will allow me to evaluate how the hits you received during sparring may have influenced your quality of life.

Please answer honestly. Your answers will be kept private.

Q1.2 Name*

Start of Block: Symptoms - Worst

Q2.1 Did you experience any of the following symptoms following Saturday's session (including Saturday afternoon)?

	Yes	No
Headache (1)	○	○
"Pressure in head" (2)	○	○
Neck pain (3)	○	○
Nausea or vomiting (4)	○	○
Dizziness (5)	○	○
Blurred vision (6)	○	○
Balance problems (7)	○	○
Sensitivity to light (8)	○	○
Sensitivity to noise (9)	○	○
Feeling slowed down (10)	○	○
Feeling like in a fog (11)	○	○
"Don't feel right" (12)	○	○
Difficulty concentrating (13)	○	○
Difficulty remembering (14)	○	○
Fatigue or low energy (15)	○	○
Confusion (16)	○	○
Drowsiness (17)	○	○
More emotional (18)	○	○
Irritability (19)	○	○
Sadness (20)	○	○
Nervous or anxious (21)	○	○
Trouble falling asleep (22)	○	○

Skip To: End of Survey If Did you experience any of the following symptoms following Saturday's session (including Saturday afternoon)? *Count of* [No] *= 22*

Carry Forward Selected Choices from "Did you experience any of the following symptoms following Saturday's session (including Saturday afternoon)?"

Q2.2 At your worst (when you experienced symptoms at the higher intensity), to what intensity did you experience these symptoms?

	0 – None	1	2 – Mild	3	4 – Moderate	5	6 - Severe
Carry Forward Selected Choices (x1-x22)	○	○	○	○	○	○	○

Q2.3 Can you remember the approximate time you felt the worst?

E.g. "4pm" or "the morning after"

Q2.4 If 100% is feeling perfectly normal, what percent of normal did you feel at your worst?*

Q2.5 Do you have any comment that could help me understand the evolution of your symptoms?

E.g. "Had a nap and felt fine again", or "Was fine for a few hours then crashed in the evening".

Start of Block: Symptoms - 48 hours

Carry Forward Selected Choices from "Did you experience any of the following symptoms following Saturday's session (including Saturday afternoon)?"

Q3.1 Let's now think about approximately 48 hours after the session (~Monday mid-day to evening): have you still been experiencing any of the symptoms below? Please rate their intensity (0 if you have not experienced it in the last few hours):

	0 – None	1	2 – Mild	3	4 – Moderate	5	6 - Severe
Carry Forward Selected Choices (x1-x22)	○	○	○	○	○	○	○

Q3.2 If 100% is feeling perfectly normal, what percent of normal do you feel right now?

Q3.3 Did your symptoms affect your <u>performance during training/sparring</u> in the last 48 hours?

Consider your ability to train at your usual intensity, to maintain your attention and energy in time, or to react to your opponent and protect yourself.

- o No effect (1)
- o To a minor extent (2)
- o To a moderate extent (3)
- o To a major extent (4)
- o I did not train in the last 48 hours (5)

Q3.4 What made you feel this way?

Q3.5 Did your symptoms affect your <u>capacity to work/study</u> in the last 48 hours?

Consider your ability to complete hard tasks, focus, handle stress and avoid distractions.

- o No effect (1)
- o To a minor extent (2)
- o To a moderate extent (3)
- o To a major extent (4)
- o I did not work/study in the last 48 hours (5)

Q3.6 What made you feel this way?

Q3.7 Did your symptoms affect your <u>personal life</u> in the last 48 hours?

Consider your ability to perform daily activities (e.g. carrying groceries, climbing flights of stairs), your emotional state and your interactions with family, friends or groups of people.

- No effect (1)
- To a minor extent (2)
- To a moderate extent (3)
- To a major extent (4)

Q3.8 What made you feel this way?

Q3.9 Did you seek medical attention related to these symptoms?

- Yes (1)
- No (3)

Q3.10 Were you diagnosed with a concussion, or mild traumatic brain injury? (conditional on Q3.10)

- Yes (1)
- No (3)

Q3.11 We are almost done. If you have any comments, please use the space below:

Prevent Hybrid Mouthguard Fitting Instructions

1. Bring water to a *rolling boil*

2. Place the mouthguard in the boiling water for EXACTLY 40 seconds
    a. Only boil **one** mouthguard at a time

3. Remove mouthguard out of the boiling water with a large spoon or ladle, let mouthguard cool under cold water for 2 seconds.
    a. Note that the mouthguard will be flimsy coming out of the boiling water, use both hands to maintain structure of the mouthguard while putting into mouth

4. Using a mirror, place and center the mouthguard on upper teeth and bite down for 10 seconds to ensure lower teeth make an indention

5. While biting down, suck in strongly for an additional 15 seconds

6. Continue to wear mouthguard for another 60 seconds and then remove the mouthguard

7. Let mouthguard cool for at least **one hour** before use.
    a. The mouthguard will have its best retention and fit once it cools.

Please note: The mouthguard will shrink as it cools. If retention is poor after the initial fitting, please let the mouthguard cool for 24 hours in order to gain retention. Do not attempt to refit the mouthguard after the initial boiling process.

End of study survey – Assessment of the comfort of the sensors

Start of Block: First page

Q1.1 Thanks again so much for participating in the study!

This is the last little bit of information I need from you. I would like some feedback about your experience with the sensors, especially the mouthguards. Please tell me what you thought of it, you can be totally honest!

Thanks again,

Enora

Q1.2 Your name*

Start of Block: Mouthguard subjective assessment

Q2.1 What type of mouthguard do you usually use?

- o Boil-and-bite (1)
- o Custom made (by a dentist) (2)
- o Other, please specify: (3) ____

Q2.2 <u>**For your usual mouthguard**</u>: please rate the following aspects:

	Extremely good (13)	Moderately good (14)	Slightly good (15)	Neither good nor bad (16)	Slightly bad (17)	Moderately bad (18)	Extremely bad (19)
General comfort (4)	o	o	o	o	o	o	o
Fit (how well it fits in your mouth, and how well it stays on your upper teeth) (5)	o	o	o	o	o	o	o
Ease of breathing ('Good' if it did not impair your breathing) (6)	o	o	o	o	o	o	o

Q2.3 <u>For the instrumented mouthguard you used for the study:</u> please rate the following aspects:

	Extremely good (13)	Moderately good (14)	Slightly good (15)	Neither good nor bad (16)	Slightly bad (17)	Moderately bad (18)	Extremely bad (19)
General comfort (4)	○	○	○	○	○	○	○
Fit (how well it fits in your mouth, and how well it stays on your upper teeth) (5)	○	○	○	○	○	○	○
Ease of breathing ('Good' if it did not impair your breathing) (6)	○	○	○	○	○	○	○

Q2.4 Do you feel like the <u>instrumented mouthguard</u> affected your overall performance?

- A great deal (11)
- A lot (12)
- A moderate amount (13)
- A little (14)
- None at all (15)

Q2.5 Would you agree to use the instrumented mouthguard again for future studies?

	Yes (1)	No (4)
For tech sparring (~Wednesday sessions) (1)	○	○
For competitive sparring (~Saturday sessions) (4)	○	○
For fighting (2)	○	○

Q2.6 Do you have any comments or additional concerns about the mouthguard?

Start of Block: Skin patch subjective assessment

Q3.1 Do you feel like the <u>patch taped behind your ear</u> affected your overall performance?

- A great deal (11)
- A lot (12)
- A moderate amount (13)
- A little (14)
- None at all (15)

Q3.2 Did you experience any discomfort while wearing the <u>patch behind your ear</u>?

Q3.3 Would you agree to use the skin patch again for future studies?

	Yes (1)	No (4)
For tech sparring (~Wednesday sessions) (1)	○	○
For competitive sparring (~Saturday sessions) (4)	○	○
For fighting (2)	○	○

Q3.4 Do you have any comments or additional concerns about the patch behind the ear?

Start of Block: Headgear patch subjective assessment

Q4.1 Do you feel like the <u>patch taped on your headgear</u> affected your overall performance?

- A great deal (11)
- A lot (12)
- A moderate amount (13)
- A little (14)
- None at all (15)

Q4.2 Did you experience any discomfort while wearing the <u>patch on your headgear</u>?

Q4.3 Would you agree to use the headgear patch again for future studies?

	Yes (1)	No (4)
For tech sparring (~Wednesday sessions) (1)	○	○
For competitive sparring (~Saturday sessions) (4)	○	○
For fighting (2)	○	○

Q4.4 Do you have any comments or additional concerns about the headgear patch?

Start of Block: Last page

Q5.1 Almost there!

At the beginning of the study, you completed a survey about your history of concussion and boxing experience/habits. This survey is also available online and being filled by combat sports athletes worldwide. Do you authorise me to copy your data (anonymously) to this larger survey to be pooled with the other responses?

- Yes (1)
- No (2)

Q5.2 If you have any last comments, please use the space below:

Q5.3 Thank you!

Appendix E - Supplementary materials for Chapters 4, 5, and 6

Data reduction flow charts (Chapters 4 and 5)

The following flow charts show the data reduction process used in Chapters 4 and 5 for the mouthguard, skin patch, and headgear patch, showing the number of sensor recordings and summary peak linear and angular accelerations at each step (PLA and PAA, respectively). The included recordings follow the left-hand side path, and the excluded recordings are shown on the right. The inclusion/exclusion criteria for the first steps (from “All recorded” to “Video-verified, true positives”) are detailed in Chapter 4, and the following steps (from “Video-verified, true positives” to “‘Good’ recordings”) are detailed in Chapter 5.

The distribution of PLA and PAA was verified with a Shapiro-Wilk test at each inclusion level and was found to be significantly skewed at all levels. Wilcoxon’s rank-sum test was therefore chosen to analyse the differences between the included and excluded subsets, and the results are reported on the flow charts. The statistics along the “Included” branch relate to the differences in median PLA and PAA between the parent and child levels; the statistics along the “Excluded” branch relate to the differences in median PLA and PAA between the included and excluded subsets.

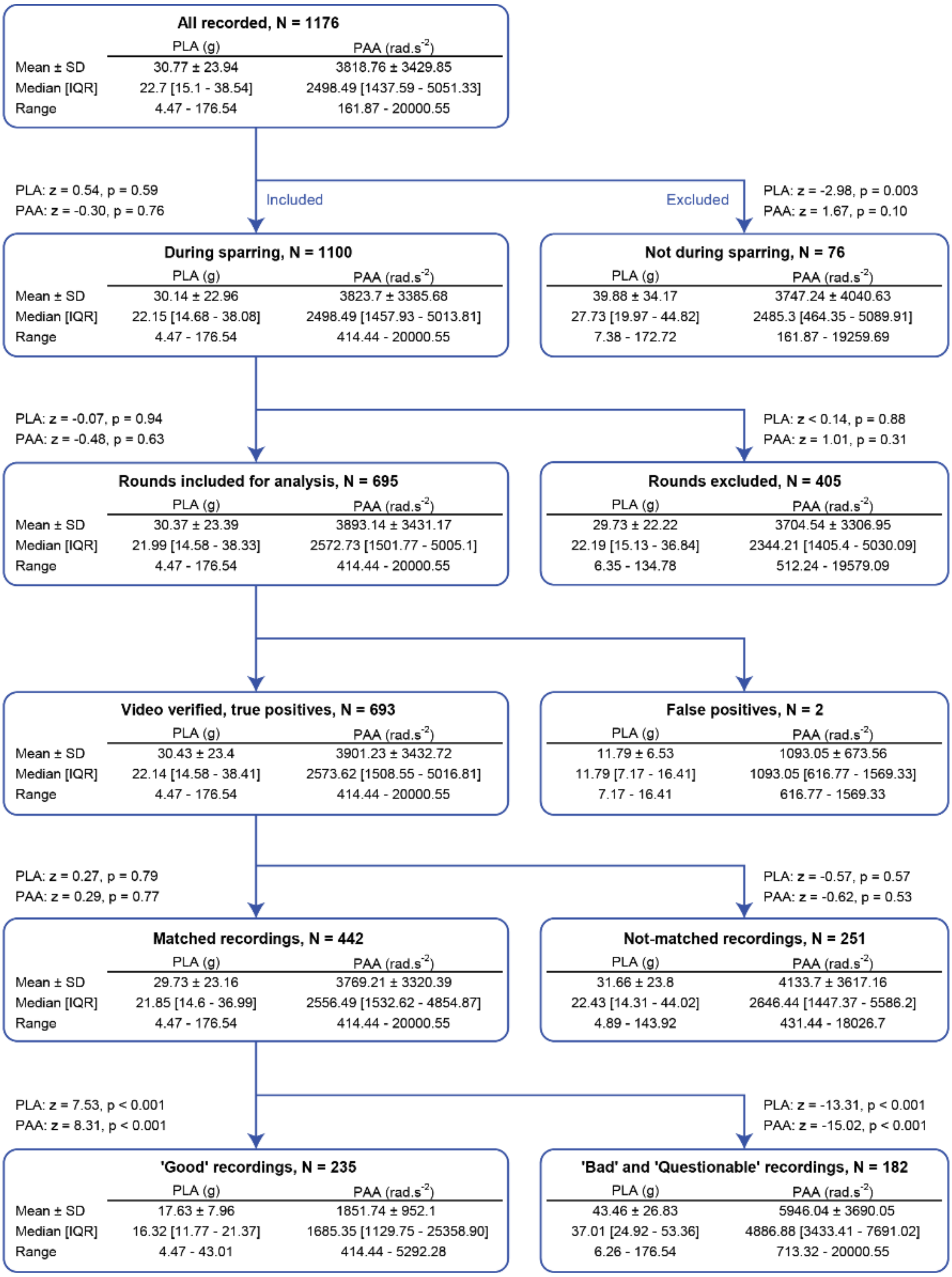


Figure E.1. Data reduction process for the mouthguard events: number of events excluded at each step and summary metrics for each subset.

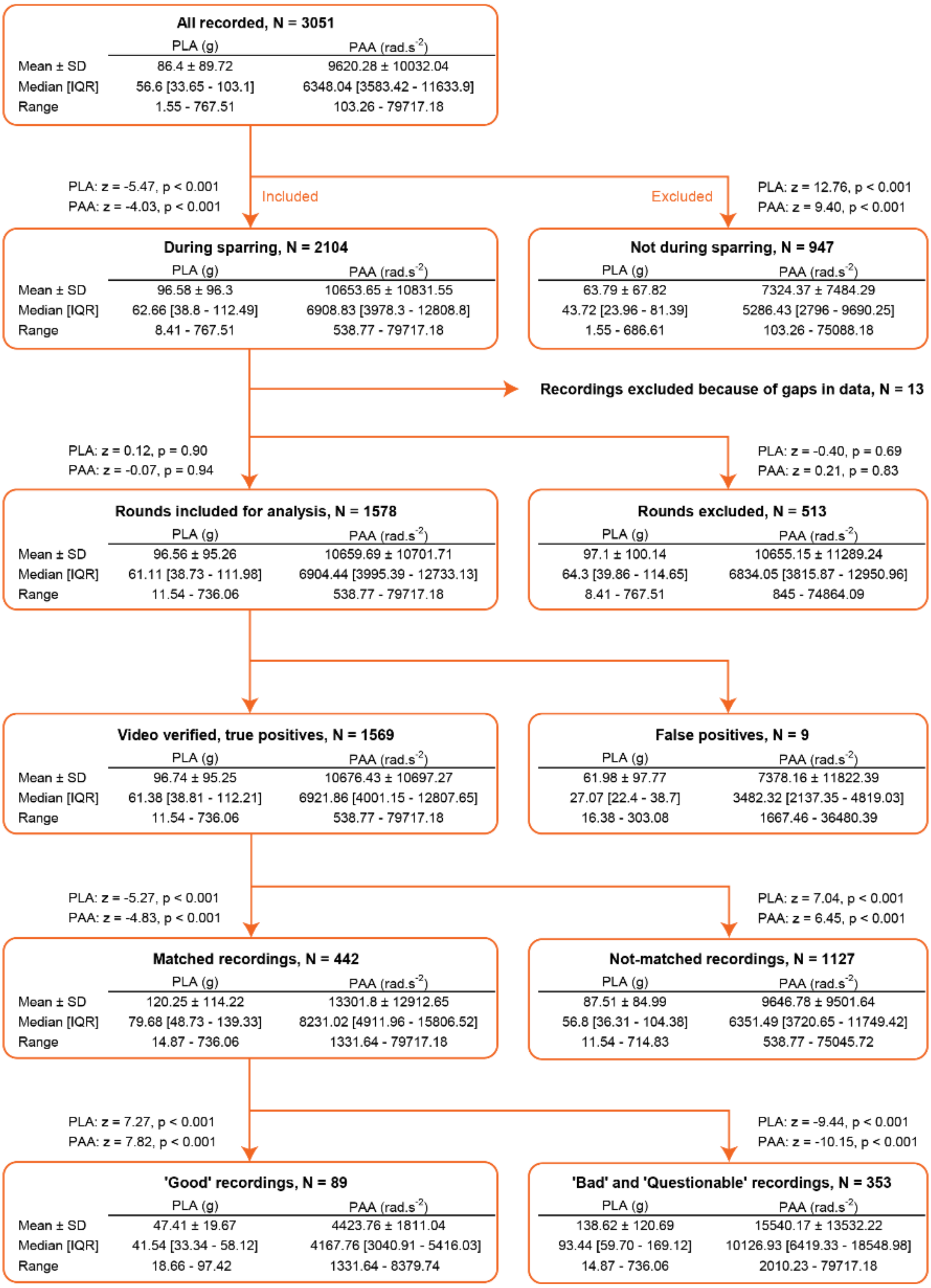


Figure E.2. Data reduction process for the skin patch events: number of events excluded at each step and summary metrics for each subset.

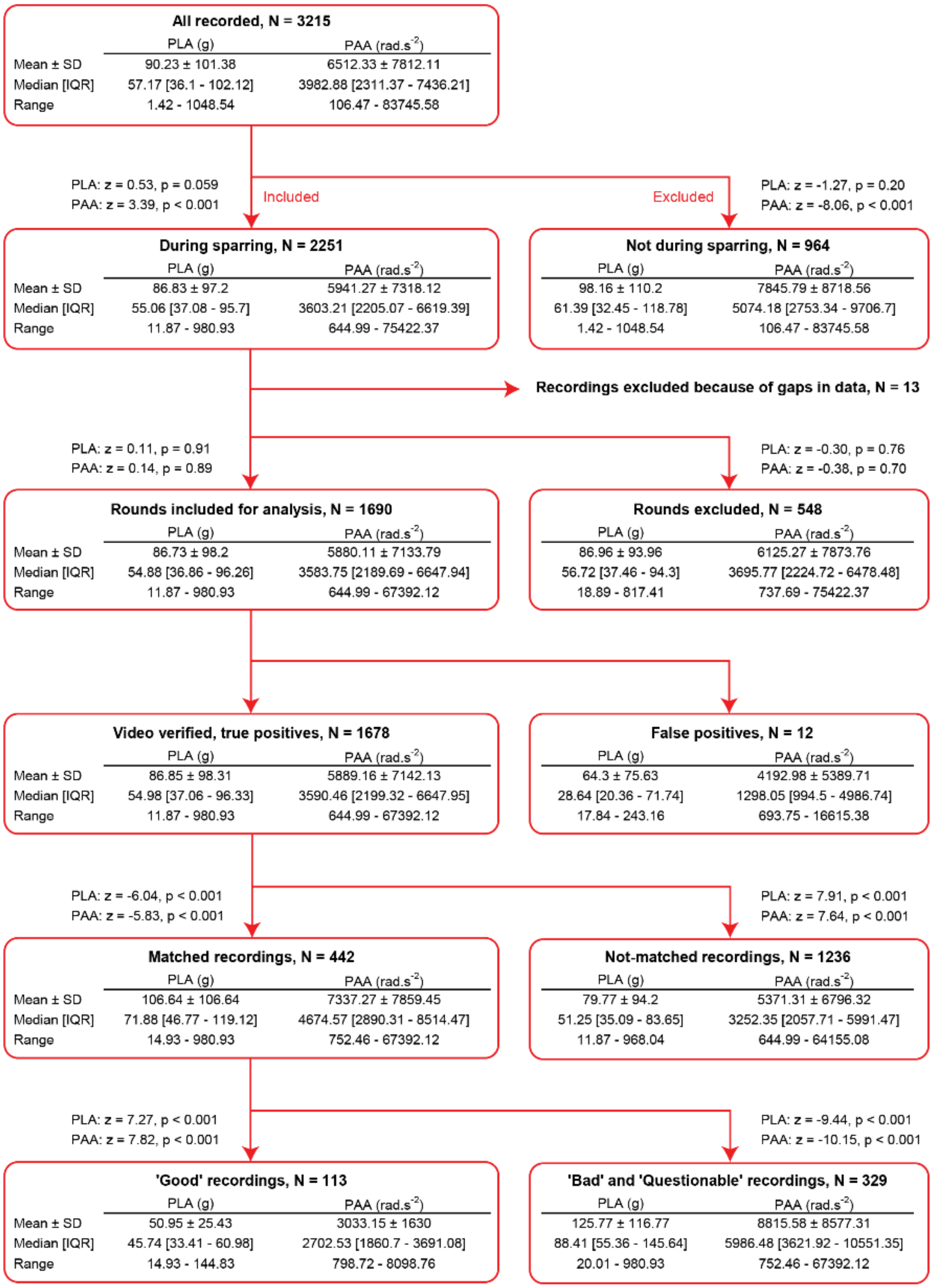


Figure E.3. Data reduction process for the headgear patch events: number of events excluded at each step and summary metrics for each subset.

Sparring-relating symptoms (Chapter 6)

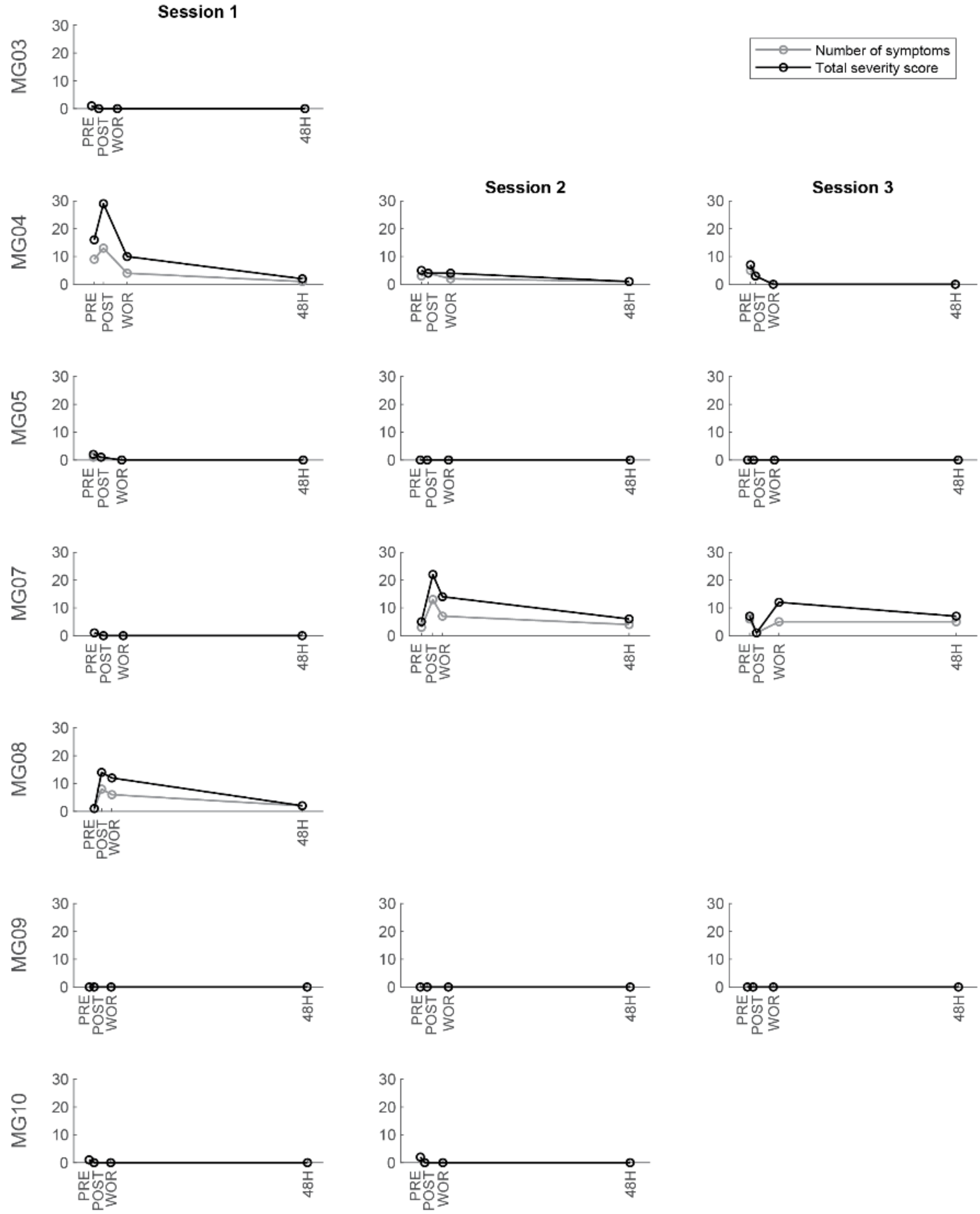


Figure E.4. Evolution of the number of symptoms and the total severity score for each athlete-session. PRE: pre-sparring, POST: immediately post-sparring, WOR: self-reported worst time where the participant experienced symptoms at the highest intensity, 48H: approximately 48 hours post-sparring.

Head impacts kinematics (Chapter 6)

Table E.1.Median and inter-quartile range [IQR] for each athlete-session and each metric, for all individual video-verified, 'good' mouthguards recordings associated with a head impact.

	Median [IQR]									
Athlete-session	PLA (g)	PAV (rad.s^{-1})	PAA (rad.s^{-2})	$\Delta\omega_{max}$ (rad.s^{-1})	GSI	HIC_{36}	RIC_{36}	BrIC	GAMBIT	HIP (W)
MG03										
Session 1	18.1 [12.9 - 23.7]	10.4 [8.6 - 13.7]	1730.1 [1219.5 - 2458.3]	10.2 [8.6 - 13.1]	5.2 [3.2 - 11.6]	5.1 [3 - 11.4]	3.7E+05 [1.9E+05 - 8.7E+05]	0.2 [0.2 - 0.3]	0.2 [0.2 - 0.3]	441.1 [214.1 - 1015.4]
MG04										
Session 1*	10.5 [8.7 - 13.8]	12.4 [10.3 - 14.9]	1216.8 [920.5 - 1718.7]	10.4 [9.1 - 12.6]	2.5 [1.4 - 4.4]	2.5 [1.7 - 4.9]	3.0E+05 [2.1E+05 - 5.4E+05]	0.2 [0.2 - 0.3]	0.2 [0.1 - 0.2]	342.2 [200.6 - 611.3]
Session 2	12 [10.1 - 32.6]	12.4 [10.1 - 15.8]	1894.5 [1638.1 - 2626.8]	11.2 [10.3 - 13.7]	2.4 [1.2 - 17.4]	2.4 [1.7 - 14.5]	5.8E+05 [3.0E+05 - 1.2E+06]	0.3 [0.2 - 0.3]	0.2 [0.2 - 0.4]	393.2 [147 - 1451.5]
Session 3	10.6 [10.1 - 14.6]	13.3 [11.2 - 15.4]	1122 [1036.7 - 1890.5]	8.1 [6.8 - 12.5]	2.4 [1.5 - 3.6]	2.2 [1.6 - 3]	2.3E+05 [1.8E+05 - 6.8E+05]	0.2 [0.2 - 0.3]	0.2 [0.1 - 0.2]	239.5 [130.1 - 409]
MG05										
Session 1	16.5 [10.8 - 24.8]	14.9 [10.5 - 18.7]	2033.6 [1285.3 - 3061.3]	13.3 [9.9 - 18.7]	5.1 [1.4 - 11.3]	5.4 [2.3 - 13.6]	9.6E+05 [3.9E+05 - 1.6E+06]	0.3 [0.2 - 0.4]	0.3 [0.2 - 0.4]	495.7 [248.5 - 1275.5]
Session 2	16.3 [11.8 - 20.2]	13.2 [10.3 - 16.9]	1762.7 [1271 - 2286.8]	12.4 [9.9 - 14.1]	5.6 [3.1 - 10.1]	4.7 [3.2 - 8.1]	4.9E+05 [3.3E+05 - 1.2E+06]	0.3 [0.2 - 0.3]	0.2 [0.2 - 0.3]	377.8 [256 - 828.5]
Session 3	11.9 [10.4 - 15.3]	11.5 [10.8 - 14.1]	1480.3 [1430 - 2353.6]	11.2 [8.9 - 13.4]	3.5 [2.3 - 5.4]	3.6 [2.4 - 4.8]	4.6E+05 [2.8E+05 - 8.1E+05]	0.2 [0.2 - 0.3]	0.2 [0.2 - 0.3]	427.7 [324.3 - 564.1]
MG07										
Session 1	17.1 [13.4 - 21.7]	14.7 [10.6 - 20.2]	1894.8 [1404.7 - 3365.2]	13.6 [11.7 - 19.3]	4.6 [2.4 - 8.2]	4.1 [3.1 - 8.5]	7.3E+05 [3.8E+05 - 1.7E+06]	0.3 [0.2 - 0.4]	0.2 [0.2 - 0.4]	424.8 [307.4 - 780.9]
Session 2*	17 [11.6 - 22.4]	14.6 [10.9 - 18.2]	1760.6 [1440.6 - 2327.1]	11.7 [9 - 14.6]	6.5 [3.2 - 9]	5.4 [3.2 - 7.8]	6.0E+05 [3.6E+05 - 9.0E+05]	0.3 [0.2 - 0.3]	0.2 [0.2 - 0.3]	407.7 [253.4 - 763.6]
Session 3*	15.3 [11.7 - 21.3]	12.6 [10 - 16]	1744.4 [1210.8 - 2119.7]	11.8 [8.9 - 14.4]	4 [2.4 - 11.2]	4.5 [2.8 - 10.1]	5.5E+05 [3.4E+05 - 8.2E+05]	0.3 [0.2 - 0.3]	0.2 [0.2 - 0.3]	455.4 [318.9 - 702.8]
MG08										
Session 1*	15 [11.4 - 19.4]	12.2 [9.9 - 17.7]	1780.2 [907.3 - 2415.1]	10.6 [8.4 - 16.2]	5.3 [2.7 - 8.9]	5 [2.9 - 9.2]	6.4E+05 [2.2E+05 - 1.0E+06]	0.2 [0.2 - 0.3]	0.2 [0.1 - 0.3]	508 [251.6 - 941.7]

* indicates the sessions where the participant reported an increase in total symptom severity score (TSSS) equal or greater than 5 between pre-sparring and any time point post-sparring.

Table E.1.(continued). Medians and inter-quartile ranges [IQR] for each athlete-session and each metric, for all individual video-verified, 'good' mouthguards recordings associated with a head impact.

	Median [IQR]									
Athlete-session	PLA (g)	PAV (rad.s^{-1})	PAA (rad.s^{-2})	$\Delta\omega_{max}$ (rad.s^{-1})	GSI	HIC_{36}	RIC_{36}	BrIC	GAMBIT	HIP (W)
MG09										
Session 1	19.6 [13.6 - 24.5]	10.6 [8.3 - 14.6]	1300.4 [982 - 2018.1]	10.2 [7.6 - 13]	9.2 [4.1 - 18.6]	8 [3.7 - 16.6]	3.1E+05 [1.4E+05 - 8.8E+05]	0.2 [0.1 - 0.3]	0.2 [0.1 - 0.3]	736.3 [381.7 - 1496.2]
Session 2	15 [11.6 - 18.3]	10.3 [7.1 - 14.3]	1116.1 [859 - 1517.5]	9.2 [6.6 - 12.2]	6.3 [3.5 - 8.5]	5.8 [3.7 - 7.4]	2.2E+05 [1.4E+05 - 3.6E+05]	0.2 [0.1 - 0.3]	0.2 [0.1 - 0.2]	457.7 [301 - 716.2]
Session 3	11.7 [9.4 - 17]	10.7 [8.5 - 13.3]	1488.2 [1062.9 - 1578.7]	9 [6.9 - 9.2]	1.8 [0.7 - 6.2]	2 [1.1 - 4.7]	3.1E+05 [1.5E+05 - 4.7E+05]	0.2 [0.2 - 0.2]	0.2 [0.2 - 0.2]	199.9 [135.9 - 498.3]
MG10										
Session 1	16.2 [11.3 - 20.9]	11.7 [9.5 - 15.7]	1416.1 [1033.6 - 1838.2]	11.6 [8.5 - 15.6]	6.5 [2.7 - 12.6]	5.9 [2.8 - 11.2]	4.6E+05 [2.1E+05 - 8.5E+05]	0.2 [0.2 - 0.3]	0.2 [0.2 - 0.2]	623 [307.6 - 1060.8]
Session 2	13.9 [10.8 - 16.5]	12.7 [9.4 - 15.4]	1126.5 [999.2 - 1668]	11.2 [7.4 - 13.8]	4.9 [1.7 - 6]	4.1 [2.4 - 5.1]	2.7E+05 [1.9E+05 - 4.5E+05]	0.2 [0.2 - 0.3]	0.2 [0.2 - 0.2]	508.3 [281.1 - 602.2]
Overall median [IQR]	15.5 [11.4 - 20.9]	12.2 [9.5 - 16]	1502.5 [1070.3 - 2120.8]	11 [8.6 - 14.5]	5.3 [2.5 - 10.1]	4.9 [2.7 - 9.2]	4.2E+05 [2.1E+05 - 8.7E+05]	0.23 [0.18 - 0.3]	0.21 [0.16 - 0.29]	482.6 [275.4 - 903.7]

* indicates the sessions where the participant reported an increase in total symptom severity score (TSSS) equal or greater than 5 between pre-sparring and any time point post-sparring.

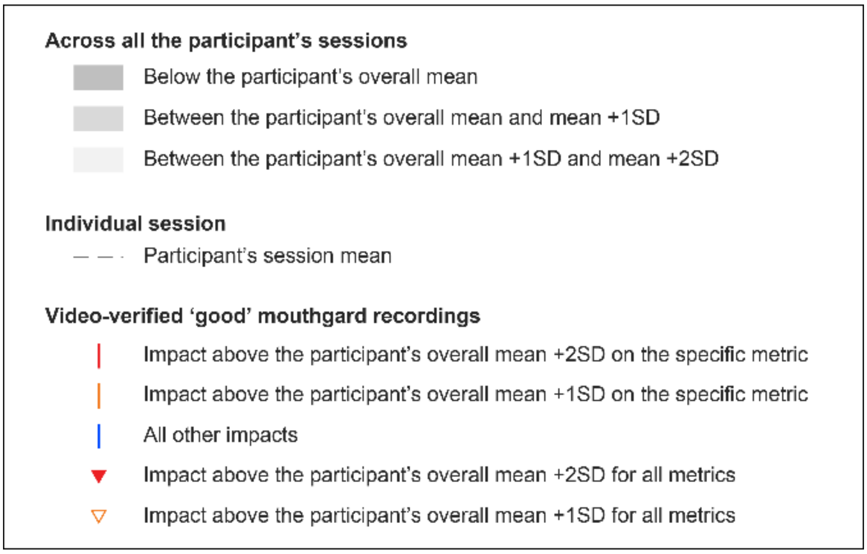
Table E.2. Means and standard deviations measured for each athlete-session and each metric, out of all individual video-verified, 'good' mouthguards recordings associated with a head impact.

	Mean ± SD								
Athlete-session	PLA (g)	PAV (rad.s^{-1})	PAA (rad.s^{-2})	$\Delta\omega_{max}$ (rad.s^{-1})	GSI	HIC_{36}	RIC_{36}	BrIC	GAMBIT	HIP (W)
MG03										
Session 1	19.1 ± 8.7	11.5 ± 4.2	2001.4 ± 1240.4	11.1 ± 4.4	9.4 ± 10	8.4 ± 8.3	9.9E+05 ± 1.7E+06	0.2 ± 0.1	0.3 ± 0.2	714.4 ± 653.4
MG04										
Session 1*	12.5 ± 6	12.9 ± 4.1	1394.9 ± 621.3	11.4 ± 4.1	4.6 ± 6.2	4.7 ± 5.7	5.0E+05 ± 6.3E+05	0.2 ± 0.1	0.2 ± 0.1	513.7 ± 542.8
Session 2	19.6 ± 13.8	13.7 ± 4.7	2429.1 ± 1480.2	12.8 ± 4.6	13.1 ± 21.2	11.9 ± 19.1	1.6E+06 ± 2.7E+06	0.3 ± 0.1	0.3 ± 0.2	798.8 ± 871.6
Session 3	12 ± 3.8	12.6 ± 3.6	1373.6 ± 575.8	9.1 ± 3.6	3.4 ± 2.8	2.9 ± 2.2	3.8E+05 ± 3.1E+05	0.2 ± 0.1	0.2 ± 0.1	351.9 ± 310.5
MG05										
Session 1	17.4 ± 8.7	14.9 ± 5.2	2252.8 ± 1190.4	14.2 ± 5.7	10.5 ± 17.2	10.2 ± 13.8	1.4E+06 ± 1.7E+06	0.3 ± 0.1	0.3 ± 0.2	918.6 ± 1081.7
Session 2	16.8 ± 7.1	13.8 ± 3.9	1830.6 ± 724.3	12.4 ± 3.2	7.8 ± 6.9	6.7 ± 5.3	7.7E+05 ± 6.1E+05	0.3 ± 0.1	0.3 ± 0.1	606.9 ± 450.9
Session 3	13.2 ± 4.4	12.8 ± 2.8	1780.4 ± 650.6	11.2 ± 2.8	5.2 ± 5.9	4.9 ± 5.3	6.3E+05 ± 5.2E+05	0.2 ± 0.1	0.2 ± 0.1	546.1 ± 507.6
MG07										
Session 1	18 ± 6.8	16 ± 6.6	2431.6 ± 1293.4	14.8 ± 5.5	8.9 ± 12.5	8.9 ± 10.7	1.6E+06 ± 2.1E+06	0.3 ± 0.1	0.3 ± 0.2	803.2 ± 851.1
Session 2*	17.1 ± 5.9	14.1 ± 4.8	1866.6 ± 605.3	12 ± 3.6	7.1 ± 4.5	6.1 ± 3.7	6.9E+05 ± 4.6E+05	0.3 ± 0.1	0.3 ± 0.1	501.6 ± 306.6
Session 3*	17.2 ± 7.9	13.5 ± 4.7	1990.9 ± 1129.4	12.2 ± 4.3	9.3 ± 12.6	8.2 ± 10.1	1.0E+06 ± 1.6E+06	0.3 ± 0.1	0.3 ± 0.1	629.9 ± 575.3
MG08										
Session 1*	15.5 ± 5.7	13.5 ± 4.4	1752.9 ± 870.4	12.1 ± 5	7.5 ± 6.7	7 ± 5.7	7.4E+05 ± 6.3E+05	0.3 ± 0.1	0.2 ± 0.1	676 ± 593
MG09										
Session 1	20 ± 8.5	12 ± 5.2	1461.7 ± 703.9	11.2 ± 5.1	13.1 ± 11.6	11.3 ± 9.8	5.3E+05 ± 5.2E+05	0.2 ± 0.1	0.2 ± 0.1	951.3 ± 719.3
Session 2	15.2 ± 5.2	11 ± 4.3	1240.6 ± 537.3	9.6 ± 3.6	6.6 ± 4.7	6 ± 3.9	3.3E+05 ± 4.1E+05	0.2 ± 0.1	0.2 ± 0.1	541.3 ± 341.2
Session 3	13.1 ± 5.2	11.5 ± 4.4	1376.6 ± 271.9	8.6 ± 1.8	3.3 ± 3.3	3 ± 2.6	3.2E+05 ± 1.7E+05	0.2 ± 0.1	0.2 ± 0	291.8 ± 229.8
MG10										
Session 1	17.2 ± 7.8	12.2 ± 4.1	1521.8 ± 674.4	12.3 ± 4.9	10 ± 10.8	8.6 ± 8.9	6.2E+05 ± 5.8E+05	0.3 ± 0.1	0.2 ± 0.1	834.2 ± 764.7
Session 2	14.3 ± 5.9	12.2 ± 4.5	1331.1 ± 519.6	11.1 ± 4.6	5.8 ± 6.1	5.2 ± 4.7	3.9E+05 ± 3.2E+05	0.2 ± 0.1	0.2 ± 0.1	544.9 ± 419.3
Overall mean ± SD	16.8 ± 7.6	12.9 ± 4.7	1728.9 ± 934.7	11.8 ± 4.7	8.7 ± 10.6	7.9 ± 8.7	7.7E+05 ± 1.1E+06	0.2 ± 0.1	0.2 ± 0.1	705.6 ± 677.7

* indicates the sessions where the participant reported an increase in total symptom severity score (TSSS) equal or greater than 5 between pre-sparring and any time point post-sparring.

Control charts (Chapter 6)

Figure E.5. (next pages) Control charts: representation of the magnitude of all video-verified 'good' mouthguard recordings collected over three full athlete-sessions for participants MG04 and MG07. Each plot shows 70 minutes and covers the full sparring session. The upper border of the shaded areas represents the participants' mean or the mean and one or two standard deviations (SD). Participant's mean and SD values were calculated over all athlete-sessions available for the participant. Each vertical line represents one recording and is blue by default. The orange and red vertical lines represent the recordings for which the magnitude was above the participant's mean +1SD (orange) or mean +2SD (red) values. The triangles highlight the recordings that were above the control limits for all 10 metrics. The asterisks by the session number (above the plots) indicate the sessions where the participant reported an increase in total symptom severity score ≥ 5 from pre- to post-sparring.



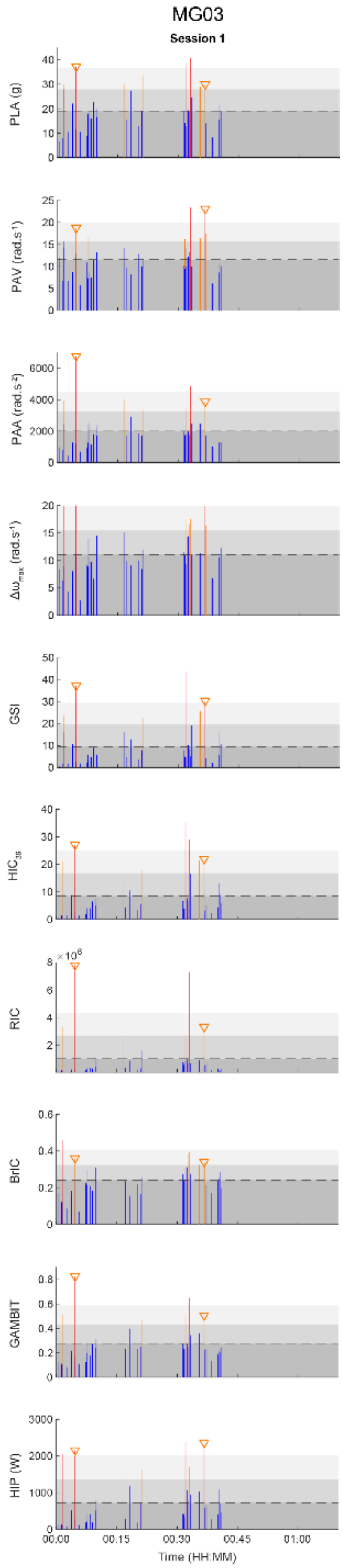


Figure E.5. (continued)

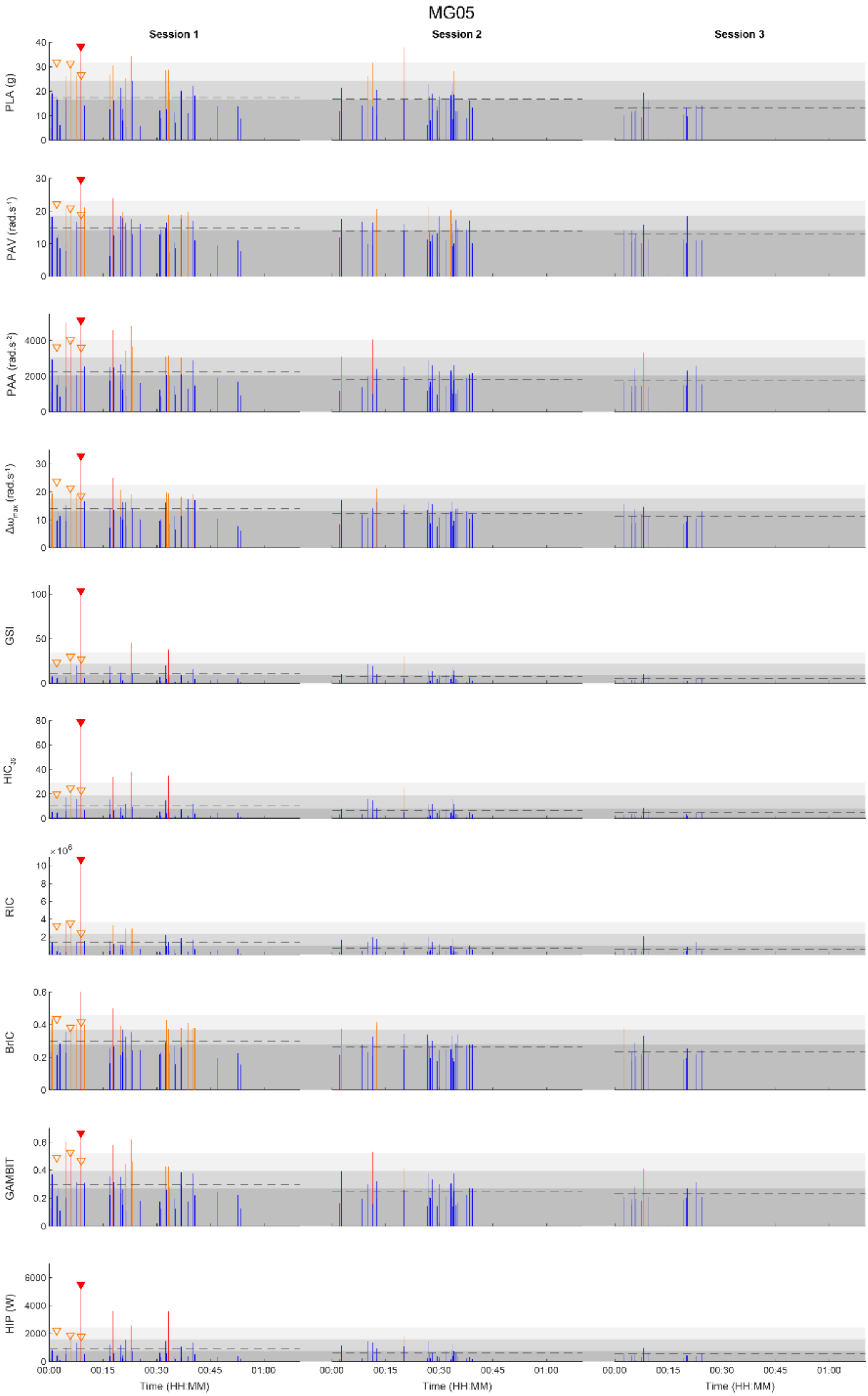


Figure E.5. (continued)

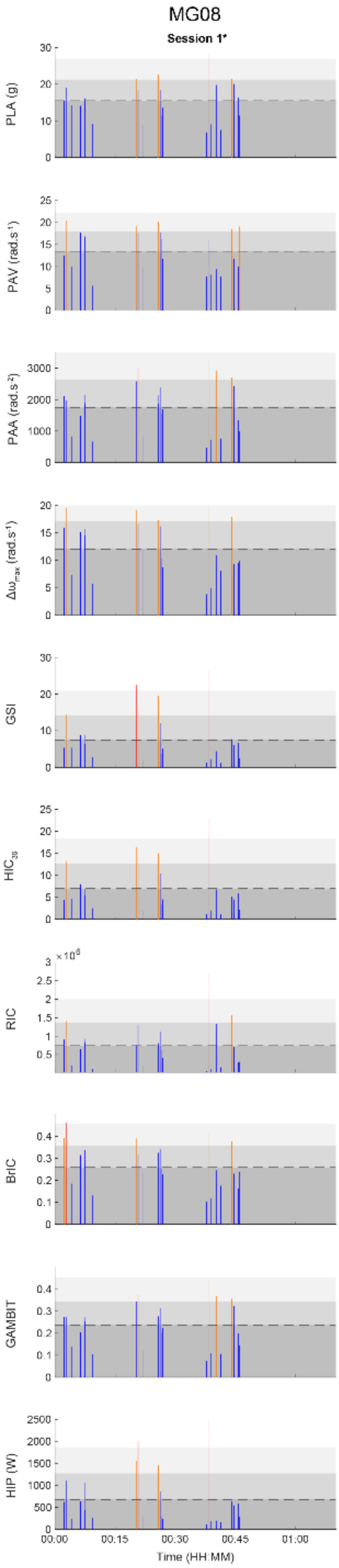


Figure E.5. (continued)

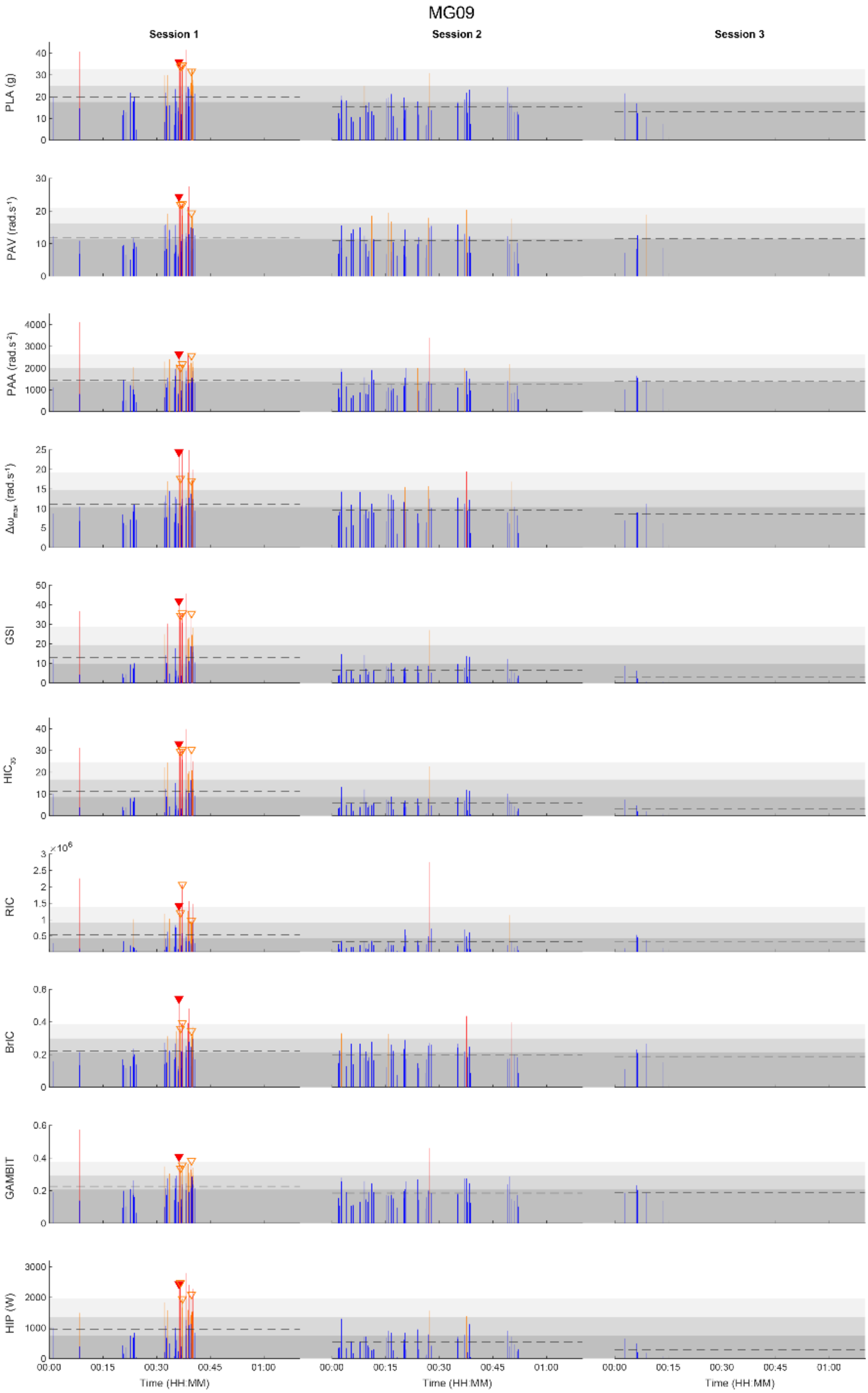


Figure E.5. (continued)

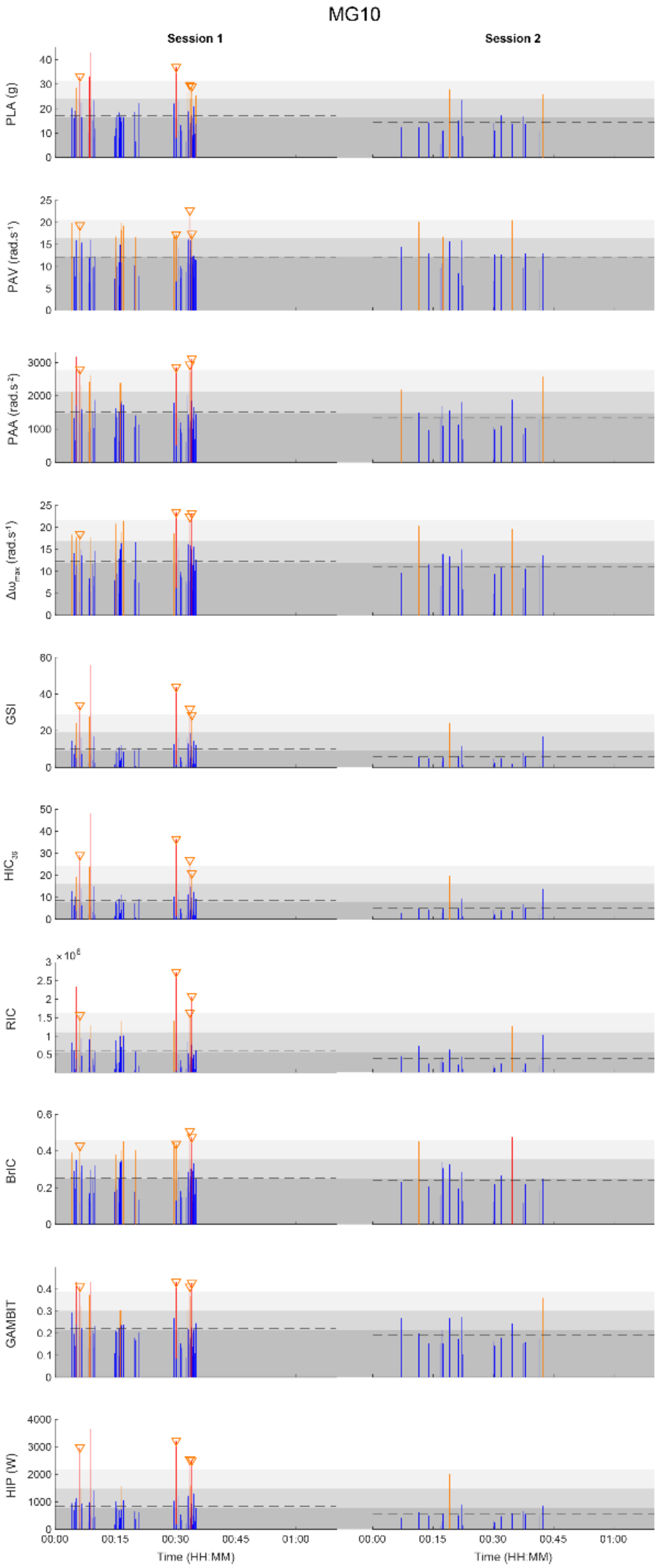


Figure E.5. (continued)

Appendix F - Evaluation of the Prevent Biometrics and CSx classification algorithm in the identification of spurious and low-quality recordings

Introduction

In Chapters 4 and 5, we evaluated the performance of the Prevent Biometrics mouthguard and CSx patch to record head impacts and measure head motion, using all events recorded by the sensors. In practice, both devices use a proprietary algorithm that determines whether a recording is representative of a true head impact (true positive) or a spurious event (false positive). End-users of these devices may only include the events that were classified as true positives by the algorithm. However, video analysis has shown for other head impact devices that between 20 and 89% of all events classified as valid by the devices proved to be false positives[67, 86, 220, 224, 258, 339, 403] and between 14 and 40% of valid impacts were improperly discarded by the algorithm.[258, 292, 322, 416]

Classification algorithms use the raw kinematic traces to determine the validity of the recording, sometimes aided by a proximity sensor, verifying the sensor is in place. The algorithms could therefore play an important role in identifying anomalies in the kinematic traces and detecting recordings that are artefacts due to the sensor's decoupling from the skull. Previous research has developed algorithms that were able to correctly identify 88-95% of artefacts measured by a skin patch, showing an improvement compared to the proprietary algorithm's performance (61%).[343]

The primary objectives of this study were to explore the performance of the Prevent Biometrics and CSx algorithms to: (1) properly classify recordings as being the result of a head or body impact (in opposition to spurious events) using video verification; and (2) distinguish high- from low-quality recordings, using the visual assessment of the raw kinematic traces' quality as the reference. Secondary objectives aimed to: (3) evaluate if participant, impact type or impact location affected the performance of the algorithm; (4) assess the capacity of the algorithm to identify the specific criteria that define low-quality recordings; and (5) determine if the mouthguard's proximity sensor readings were associated with the quality of the recordings.

This study is not part of the main body of doctoral work because of the rapidly-changing landscape of these algorithms. We reason that not only research teams,[343, 416] but also manufacturers are constantly working on modifying the algorithms, and end-users may not be aware of updates. As a result, what we investigate here may already be obsolete.

Methods

The data collection and processing methods are detailed in full in Chapters 4 and 5, along with the full description of the Prevent Biometrics mouthguard and the CSx patches.

Sensors and algorithms

Both the Prevent Biometrics mouthguard and the CSx patches incorporate a proprietary classification algorithm, providing a classification of whether the acceleration event is deemed a true head impact or a spurious recording. The characteristics of both systems are presented in Table F.1.

Table F.1. Characteristics of the devices' systems for the classification of true/false-positive events.

	Prevent Biometrics mouthguard	CSx Patch
Allows access to both true and false positive events	Yes	Yes
Availability of the classification result	Available by default on a summary spreadsheet when recordings are downloaded from the Prevent Biometrics portal	Embedded in each recording (CSV format)
Classification algorithm process	Assesses the frequency content and the signal-to-noise ratio in the signals.[198] Details are not available.	Searches for instances where the resultant LA rises then falls with a slope steeper than 10 g per sample within a 1.56-ms window (5 samples).
Proximity sensor	Yes, to estimate how close the mouthguard is to the teeth. The proximity sensor is sampled every 100 ms and gives a pre- and a post-event reading for each recording.	No
Availability of the proximity sensor reading	Not by default, but was provided to us for this study.	Not applicable

LA: linear acceleration

Analysis

We first assessed the overall performance of the classification algorithms by calculating the sensitivity, specificity, positive and negative values (PPV and NPV, respectively) and the diagnostic accuracy (Figure 4.7, page 60) from all the recordings that were identified in Chapter 4, i.e., 695, 1578, and 1690 recordings for the mouthguard, skin patch, and headgear patch, respectively. These calculations used video verification as the reference.

Next, the recordings associated with video-verified definite head impacts and prolonged contacts were used for the analysis of the false negative rates (FNR)(645, 1400, and 1498 events from the mouthguard, skin patch, and headgear patch, respectively). The FNR represents the ratio of true events incorrectly classified as spurious on all events recorded, and was calculated, along with its 95% confidence intervals (CIs), for various impact characteristics: participant, type of impact (definite head impact vs. prolonged contact, direct vs. indirect), and impact location (in the vicinity of the sensor or not, in the face area or not, by location bins). A lower FNR was associated with better performance, and a significant difference in FNR between conditions was assumed when the 95% CIs did not overlap.

The performance of the classification algorithms to identify recordings of 'good' or 'bad' quality (as defined in Chapter 5, page 86) was then assessed on a selection of events. These events were all associated with a specific definite head impact or prolonged contact verified on video, had matching recordings on all three sensors, and were assessed as being of 'good' or 'bad' quality. Recordings assessed as 'Questionable' were excluded as their quality was uncertain. The final numbers of events were 392, 386, and 292 for the mouthguard, skin patch, and headgear patch, respectively. The sensitivity, specificity, PPV, NPV, and diagnostic accuracy were calculated with their 95% CIs, using the visual quality assessment as the reference.

Next, the performance of the classification algorithm to identify the criteria defining a recording as 'bad' were examined via the calculation of the algorithms' specificity and its 95% CI for each criterion defined in Table 5.1 (page 86). A higher specificity meant that the algorithm was better at identifying the anomaly. This analysis was conducted for the 'bad' recordings only and used 157, 297, and 179 recordings from the mouthguard, skin patch, and headgear patch, respectively.

Finally, we qualitatively explored the mouthguard's proximity sensor readings, appraising if they were associated with the visual assessment of the recordings' quality. This analysis was conducted on the 442 events that were included in Chapter 5's analyses.

Results

Sensitivity and specificity of the classification algorithms

Most of the video-verified impacts were correctly predicted as true positives by the Prevent Biometrics and CSx classification algorithms, resulting in sensitivities of 83%, 91%, and 98% for the mouthguard, skin patch, and headgear patch, respectively (Table F.2). However, none of the recordings that could not be video-verified was properly classified as spurious by the sensors' proprietary algorithms.

Table F.2. Frequency tables and diagnostic accuracy metrics for the Prevent Biometrics and CSx classification algorithm to identify the recordings associated with an impact observed on video ("Valid") from the spurious recordings.

		Video verification					
		Mouthguard		Skin patch		Headgear patch	
		Valid	Spurious	Valid	Spurious	Valid	Spurious
Classification algorithm prediction	Valid	577	2	1433	9	1638	12
	Spurious	116	0	136	0	40	0
Sensitivity		83% (80% - 86%)		91% (90% - 93%)		98% (97% - 98%)	
Specificity		0% (0% - 1%)		0% (0% - 0%)		0% (0% - 0%)	
PPV		100% (99% - 100%)		99% (99% - 100%)		99% (99% - 100%)	
NPV		0% (0% - 1%)		0% (0% - 0%)		0% (0% - 0%)	
Diagnostic accuracy		83% (80% - 86%)		91% (89% - 92%)		97% (96% - 98%)	

PPV: positive predictive value, NPV: negative predictive value.

For the video-verified definite head impacts and prolonged contacts to the head specifically, the false negative rate was 16% (95%CI: 13% – 19%) for the mouthguard, 9% (95%CI: 8% – 11%) for the skin patch, and 2% (95%CI: 2% – 3%) for the headgear patch (Figure F.1). The Prevent Biometrics algorithm's performance was lower for direct impacts (FNR: 25%, 95%CI: 20 – 31% vs. indirect impacts: 11%, 95%CI: 8% – 14%) and impacts in the proximity of the sensor or to the face (24% - 28% vs. away from the sensor and the face: 13%, 95%CI: 10% – 16%). The CSx algorithm performance was strongly affected by the location of the impact, with impacts in the vicinity of the sensor and generally to the right side and back of the head for the skin patch, and to the back of the head for the headgear patch leading to worsened FNR (FNR for impacts in the vicinity of the sensors: 39% - 45%, vs. away from the sensor: 1% - 6%).

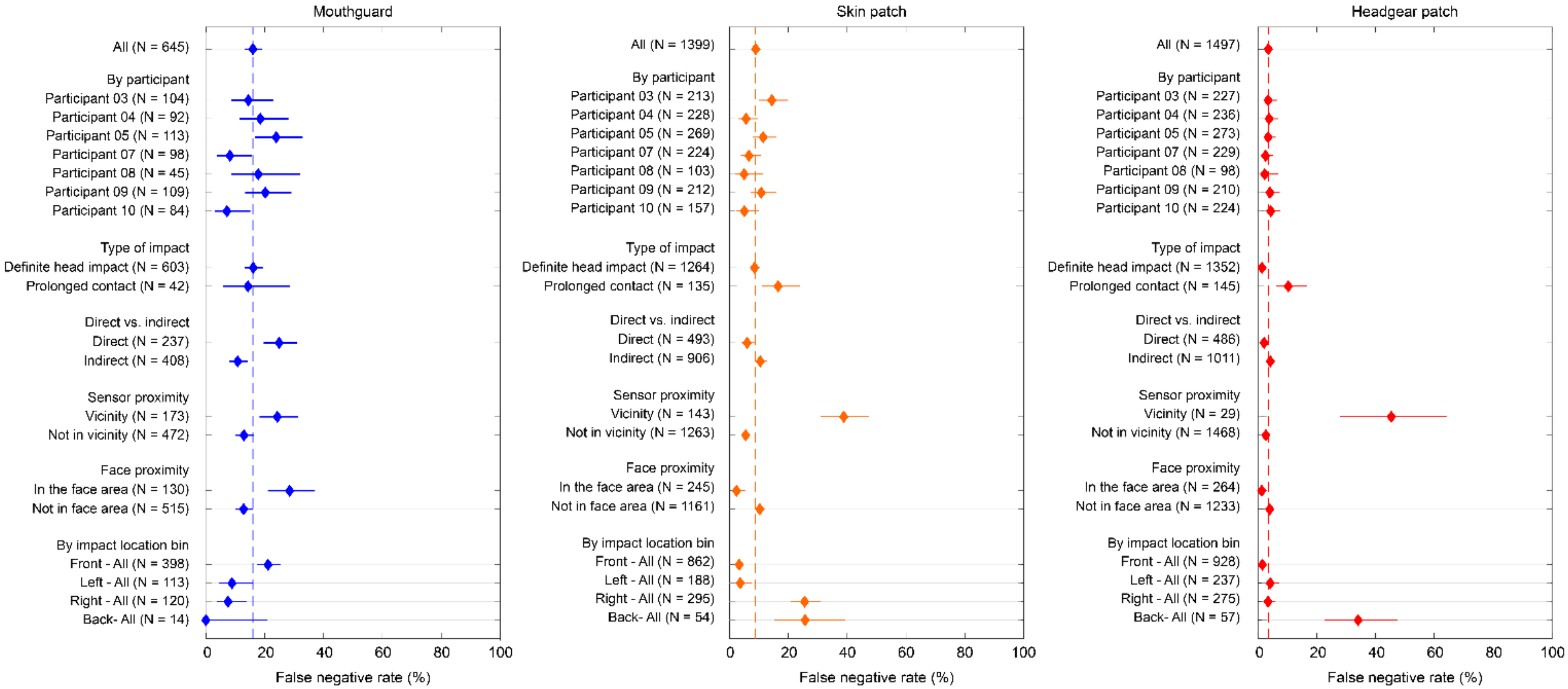


Figure F.1. Comparison of the classification algorithms' false negative rates for the mouthguard (left), skin patch (middle), and headgear patch (right) for various characteristics. The data included only video-verified definite head impacts and prolonged contacts to the head. The diamonds represent the central value of the false negative rate, the lateral bars the 95% confidence interval, and the coloured dashed vertical line represent the overall false negative rate for the sensor. A low value for the false negative rate indicates the algorithm was performing well.

Identification of low-quality recordings

Overall, the majority of the 'good' recordings were appropriately predicted as valid impacts by the Prevent Biometrics and CSx classification algorithms (sensitivity: 91-100%)(Table F.3). However, the majority of 'bad' recordings were also classified as valid (specificity: 8-24%). All the recordings predicted as spurious by the CSx algorithm, were assessed as being of poor quality (NPV of 100%) while the Prevent Biometrics algorithm classified some recordings as spurious while they were of good quality and video-verified (NPV of 65%).

Table F.3. Frequency tables and diagnostic accuracy metrics for the Prevent Biometrics and CSx classification algorithm to identify the recordings assessed as 'Good' or 'Bad' from the visual assessment.

		Visual assessment					
		Mouthguard		Skin patch		Headgear patch	
		'Good'	'Bad'	'Good'	'Bad'	'Good'	'Bad'
Classification algorithm prediction	Valid	215	120	89	235	113	165
	Invalid	20	37	0	62	0	14
Sensitivity		91% (88 - 94%)		100% (99 – 100%)		100% (98 – 100%)	
Specificity		24% (20 – 28%)		21% (17 – 25%)		8% (5 – 12%)	
PPV		64% (59 – 69%)		27% (23 – 32%)		41% (35 – 47%)	
NPV		65% (60 – 70%)		100% (99 – 100%)		100% (98 – 100%)	
Diagnostic accuracy		64% (59 – 69%)		39% (34 – 44%)		43% (38 – 49%)	

PPV: positive predictive value, NPV: negative predictive value.

The analysis of the specificity by quality criteria did not reveal any specific criterion that the Prevent Biometrics algorithm was significantly better or worse at identifying (Figure F.2). The CSx algorithm was particularly efficient at identifying the recordings where the sensors saturated (specificity of 100%), and the LA signals that presented a rapid signal inversion (skin patch only: 76%).

The proximity sensor readings (pre-, post-trigger and difference pre/post) did not explain the distribution of 'good', 'questionable' or 'bad' mouthguards recordings (Figure F.3).

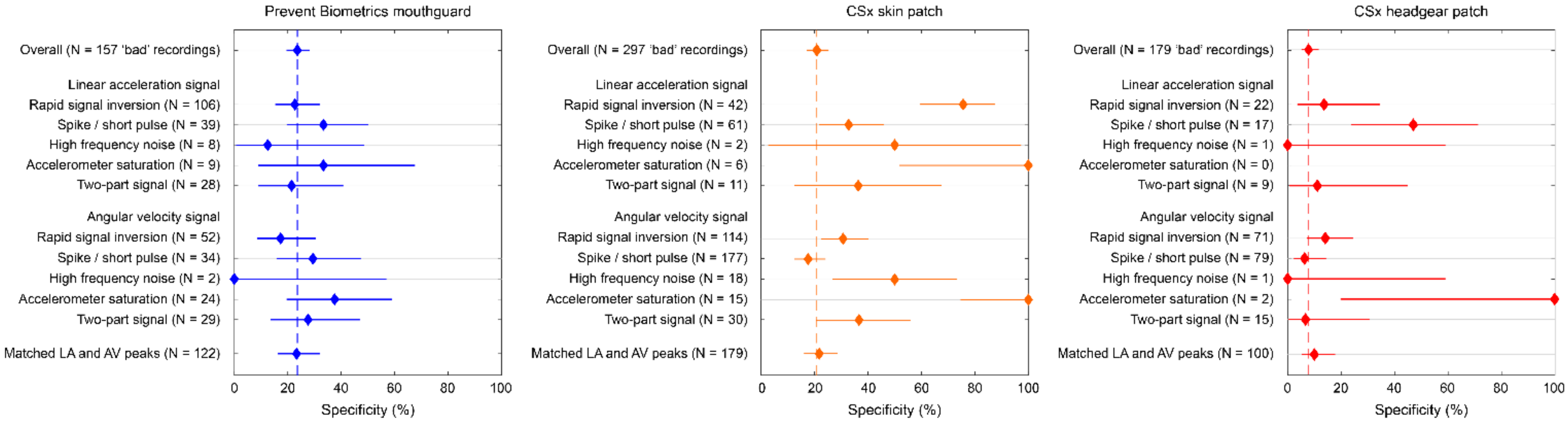


Figure F.2. Variation of the specificity of the classification algorithms for each criterion used to visually assess the recordings as 'bad'. LA: linear acceleration; AV: angular velocity.

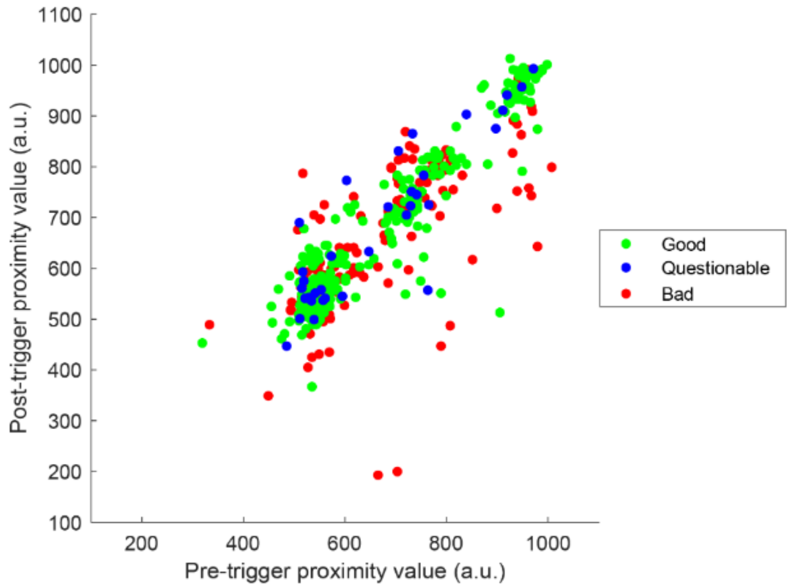


Figure F.3. Distribution of 'good', 'questionable' and 'bad' mouthguard recordings relative to the proximity sensor's reading pre- and post-trigger (N = 442, including data from seven participants). a.u.: arbitrary units. A higher value is associated with a shorter distance between the mouthguard and the teeth, thus with a theoretical better fit.

Discussion

Head impacts vs. spurious recordings

The Prevent Biometrics and CSx algorithms both classified a non-negligible proportion of recordings as spurious although they were matched to a head impact (16, 9 and 2% for the mouthguard, skin and headgear patch, respectively). Therefore, if one were to only account for the recordings predicted as valid by the devices, the number of head impacts sustained would be under-estimated.

In addition to the moderate proportions of incorrectly classified impacts, we also observed that both the Prevent Biometrics and CSx classification algorithms were sensitive to the location of the impact. Punches near the sensor (all sensors), to the face (for the mouthguard), to the right side and back (for the skin patch) and to the back of the head (for the headgear patch) all showed an increase in the rate of impacts being classified as spurious by the algorithms. This sensitivity suggests that impacts occurring in the vicinity of the sensor present acceleration and/or velocity time traces that fool the algorithm into thinking these are not valid head impacts. We verified this hypothesis in Chapter 5, showing that impacts in the vicinity of the sensor more often produced anomalies (Figure 5.3).

Quality classification

Therefore, an important strength of the classification algorithms would be to be able to identify these anomalies and differentiate between 'good' and 'bad' signals. Both algorithms classified 'good' signals as valid in most cases (sensitivity: 91% - 100%) but their capacity to identify 'bad' signals as invalid was low (specificity: 8%- 24%). A specificity of 24% means that 3 out of 4 'bad' recordings would be included in the analysis, increasing the summary PLA and PAA and introducing differences that may not exist, as seen in Chapter 5. Overall, if a user were to use the current algorithms to exclude low-quality events, they would incorrectly include 31%, 61%, and 57% of 'bad' recordings with the mouthguard, skin patch, and headgear patch, respectively. With the mouthguard, they would also incorrectly exclude 5% of 'good' recordings (0% for the patches). While our methods for visually assessing the quality of the signals have limitations (see Chapter 5's discussion), we would recommend using such methods over the devices' algorithms.

Our analysis of the algorithms' performance with respect to the quality criteria identified characteristics that could help the development of new algorithms dedicated to classifying recordings that contain good or bad data. For example, the CSx algorithm's design (Table F.1) was effective at identifying particularly sharp spikes but was affected by the general smoothness of the signal (our visual analysis highlighted that the headgear patch measurements were generally smoother than the skin patch's). The CSx algorithm identified all recordings where the sensors reached their saturation point, suggesting this may have been specifically implemented in the algorithm's design. In contrast, the Prevent Biometrics algorithm classified as valid more than 60% of the saturated recordings. It is improbable that a well-coupled sensor would measure linear accelerations over 200 g during sports participation, and the sensor reaching saturation is likely a consequence of the sensor moving independently of the skull.

The Prevent Biometrics mouthguard's proximity sensor could ideally be used to determine if the mouthguard is properly coupled to the dentition at the time of impact, which could help to identify the recordings that are not valid (i.e., recordings of sensor motion rather than head motion). Our findings showed that the current implementation of the proximity sensor's measurement did not assist in identifying low-quality signals. However, this is limited by the low sampling frequency (10 Hz) and the variable delay between an impact and the pre-/post- impact proximity measurements. Continuous sampling at a higher frequency, which is being implemented by Prevent Biometrics, may be more helpful at identifying if the mouthguard is in place, and if/when it is knocked loose by the impact. The proximity sensor could help in investigating mouthguard fit and how the fit affects the overall quality of a dataset.

Implications

While the kinematics of recordings that show the sensor has decoupled from the skull should not be used to estimate the magnitude of a head impact, it should still be counted as an exposure to a head impact (assuming it was video-verified). Therefore, an adequate classifier would be able to differentiate between three conditions: i) a head impact with good data representative of the sudden change of head velocity, ii) a head impact with unreliable data, that should be used in the impact count but not in the analysis of magnitude, and iii) an artefact that is not associated with a head impact and should be excluded from all analyses. In the future, it would be beneficial to assess the performance of a classification algorithm against these three criteria.

Appropriate classification of acceleration events is essential to get valid estimates of exposure to head impacts. A reliable algorithm could significantly reduce data processing times while approaching reference standards of video-verification and quality assessment. Machine-learning based algorithms appear particularly promising. Wu et al.[416] demonstrated that a support vector machine (SVM) algorithm resulted in sensitivity, specificity, PPV and NPV all above 87% in classifying head impacts from spurious recordings. Related to signal quality, Rooks et al.[343] showed that skin patch data could be classified between 'good' and 'bad' by a decision tree algorithm with moderate sensitivity (81%) but higher specificity (89%) than what we observed in the current study (21%). Such algorithms are expected to be adopted by head impact devices manufacturers, and their performance will need to be assessed.

In Chapter 4, we noted that the headgear patch's capacity to properly classify impacts to the head was better than the skin patch's (false negative rate: 2%, 95%CI: 2-3% vs. 9%, 95%CI: 8–11%). In the present study, we observed that the headgear patch's capacity to eliminate problematic recordings is poorer than the skin patch's capacity, and the visual assessment of the raw data showed wider pulses for the headgear patch's data than for the skin patch's data. Therefore, it is possible that the equipment and the mounting on the head, which results in different kinematic characteristics, affect the performance of the classification algorithm. As the algorithms improve, a question related to sport specificity arises: can one universal algorithm be used for all sports, or does it need to be sport-specific (in addition to sensor-specific) to perform well? For example, would there be differences in the performance of the algorithms between American football (or any hard-shell helmeted sport) and boxing (that uses padded gloves and padded headgear)?

Limitations

The classification algorithms were primarily developed to identify acceleration events that are not associated with head impacts (e.g., manipulation of the sensor).[419] In the second part of this study, we assessed the performance of the classification algorithms to identify anomalies in signals that were associated with head impacts. We cannot assume that the algorithms were programmed to assess the quality of the data as we defined it. The kinematics of non-head impacts and of 'bad' head impacts may present different characteristics that the algorithms were not developed to identify. It is also possible that these algorithms have been developed using datasets that contained 'bad' impacts that had been labelled as valid because it was assumed that all head-impact-related acceleration events were to be included. Consequently, the accuracy values reported here should be interpreted cautiously, and with the intent to better understand the signals anomalies and how algorithms could be used for their identification. Future work should consider classifying impacts into more than two categories so that users could better discriminate those impacts that should and should not be included in calculations focused on the number and magnitude of head impacts.

We only analysed the performance of the algorithms for recordings triggered during sparring rounds, which did not include recordings triggered by the manipulation and handling of the sensors. Therefore, the overall performance of the sensors' algorithm may be different from what we report here. Our results also cannot be compared with other studies that included all recordings.[258, 292, 322, 416] We expect that both algorithms would show a higher specificity as they would exclude manipulation-related recordings that typically present different frequency components.[17, 419] Specifically, the Prevent Biometrics mouthguard's algorithm was designed to assess the frequency content of the signals,[198] and its proximity sensor would have helped in detecting the recordings triggered when the mouthguard was not in the mouth.

Conclusions

Our findings highlighted that the sensors' classification algorithms were sensitive to issues related to skull/sensor decoupling for video-verified head impacts. Given the overall large proportion of 'bad' recordings in our data, there is a strong potential for algorithms to assist in the classification of acceleration events and to differentiate between valid head impact measurements, head impacts measurements resulting from skull/sensor decoupling and non-head impacts artefacts.

Appendix G - Conference abstract: Head impact monitoring: What new methodologies could do for concussion biomechanics

This appendix was presented at the 2018 conference of the International Society of Biomechanics in Sports.

> Le Flao, E., Hume, P., & King, D. (2018). Head impact monitoring: what new methodologies could do for concussion biomechanics. *Proceedings of the 36th International Conference of Biomechanics in Sports.,* 36(1), 1041-1044.

Introduction

Concussion has become a world-wide concern for collision and contact sports participants.[256] It is one of the most commonly reported injuries in sports such as American Football[112] or rugby.[132] Recent research has revealed that a history of concussions is associated with long-term health impairments.[171] Concussions are defined as a mild subset of traumatic brain injuries, and typically occur following a direct impact to the head or an impact elsewhere on the body with an "impulsive" force transmitted to the head.[256] The forces resulting from such impacts are conveyed to the brain and generate internal strains that have been proposed as the main injury mechanisms.[168] Current technology does not yet facilitate the measurement of brain strains *in-vivo*, so devices measuring real-time head kinematics have been developed.[115, 209] Acceleration instrumented devices have allowed the collection of thousands of head impacts, with the hope of finding injury thresholds that enable the identification or prevention of concussions.

The aim of this research was to capitalize on fifteen years of head impact monitoring, describe results, expose limitations, and explore directions for future research.

Methods

A narrative review was first conducted to provide an overview of the current state of knowledge around head impact biomechanics in sports-related concussion research. Subsequently, a systematic review was performed to specifically assess prospective cohort studies that reported *in-vivo* head impact monitoring in sports. The search utilised the keywords: ((head injur*) OR (brain injur*) OR concuss*) AND ((head impact*) OR acceler* OR biomechanic*). Articles were excluded if: (i) full text was unavailable in English or French; (ii) no head impact measurement was performed; (iii) they were a review, laboratory study, case report or case series, commentary or opinion piece; (iv) were not peer-reviewed. For all publications that met the inclusion and exclusion criteria, information was extracted: population details, technology utilised, number of impacts and of concussions recorded, reporting of head kinematics and effects of various factors on head kinematics.

Results & discussion

A total of 120 articles were retained for the narrative review, and 129 articles met the inclusion and exclusion criteria of the systematic review. The annual number of publications reporting head kinematics has steadily increased in the previous fifteen years, from 1 in 2005[115] when the Head Impact Telemetry System (HITS) was first utilised, to 36 in 2017. The majority of the research (81%) was conducted in helmeted sports, mainly American Football (55%) and ice hockey (13%), using the HITS. Custom devices were seldomly utilised until the first commercially available instrumented mouthguard[209] and ear patch,[257] which led to a 5-fold increase in the number of studies on non-helmeted sports in the following 3 years.

The results presented in this section are focused on male athletes of high school age or older (≥ 15 years)(N = 67 studies) to allow comparison with previously published studies.[38] The most commonly reported variables were the resultant peak head linear acceleration (PLA) and the resultant peak head angular acceleration (PAA)(Table G.1). The mean values for 50th and 95th percentiles of PLA and PAA are presented in Table G.1. Attempts have been made to determine concussion injury risk thresholds based on concussive and non-concussive impacts reports. A recent meta-analysis showed that the average values of PLA and PAA for concussed male athletes (high school level and older) were 98.7 g and 5776.6 rad.s^{-2} [38] (Table G.1). However, concussions in adult male athletes have been associated with impacts as low as than 41g and 3,000 rad.s^{-2}.[229, 250] On the opposite side of the spectrum, studies[261, 378] reporting PLA and PAA values as high as 180 g and 16,000 rad.s^{-2} did not report any diagnosed concussions. As a result of this variability, the hopes of finding injury thresholds using these single variables were left unfulfilled.

Table G.1 Reporting of the main head impact metrics, overall mean (SD) for the 50th and 95th percentiles and reported mean for concussive impacts for male athletes (≥ 15 years).[38]

Reported variable	Percentage of studies reporting (N = 67)	Overall 50th percentile (SD)	Overall 95th percentile (SD)	Mean of concussive impacts (95% CI)
PLA (g)	84%	19.2 (4.1)	51.7 (8.0)	98.7 (82.4 – 115.0)
PAA (rad.s^{-2})	67%	1543 (1270)	4099 (2632)	5776 (4584 – 6969)

SD: Standard deviation; PLA: Peak linear acceleration; PAA: Peak angular acceleration. Overall means are calculated for the studies reporting summary values for the whole cohort, including concussive and non-concussive impacts.

A multitude of factors might explain why the association between head impact biomechanics and concussion remain elusive. Firstly, the diagnostic of concussion itself is imperfect, as it currently relies on injured players reporting their symptoms,[303] unreliable tools,[88] and a disputed injury definition.[88] Secondly, several intrinsic risk factors may play an important role in the variability of the accelerations observed in injured players. The only intrinsic risk factor to have been validated with a high level of certainty was the history of concussions.[2] Other potential aspects to consider include neck strength and pre-injury symptoms. Finally, the measurement and analysis of head kinematics presents numerous limitations.

Head impact devices come with measurement errors, requiring an informed interpretation of the results.[369] The most commonly used system, the HITS, presents a technology different to most of the other devices which are typically one unit composed of a triaxial accelerometer and gyroscope. The HITS consists of six single-axis accelerometers recording linear acceleration and uses proprietary algorithms to provide PLA, PAA, the location of the impact, and several injury metrics described below. Important limitations of the HITS may reside in the fact that the algorithms are unavailable to the end user[4] and that angular accelerations are estimated from linear acceleration, using an assumed fixed head rotation point and an approximated head centre of gravity.[369]

Historically, PLA and PAA have been analysed separately but composite injury metrics have been utilised or developed. The Head Injury Criterion (HIC) uses PLA and an arbitrary time interval.[402] The Head Impact Telemetry severity profile (HITsp) weights PLA and PAA, HIC, GSI and impact location.[147]

As shown in Table G.1, head impacts are highly skewed towards the lowest acceleration magnitudes, but sports participants can sustain a high number of head impacts per game[209] and the effects of repetitive non-concussive impacts on the occurrence of injury remain unknown. Researchers have developed a cumulative metrics to account for the number or impacts and their magnitude,[120, 397] or the time between successive impacts preceding an injury.[43] However, the accuracy of these metrics in classifying head impacts as concussive or not has not yet been fully investigated.

Supposedly because head impact devices typically provide single, discrete values, such as PLA and PAA, the analysis of the acceleration signals has been severely restricted. As biomechanists, we understand that an acceleration signal is not limited to a set of discrete values from resultant curves but is composed of six time series for translation and rotation along the three axes of motion. This whole raw signal is now available from several custom as well as commercially available head impact devices, allowing a comprehensive analysis of head kinematics. Various methods of increasing popularity, such as principal component analysis, waveform analysis or machine learning, are now offered to sport scientists. These methods would most likely provide a new understanding of head impacts, and therefore a new understanding of concussion. By analysing other aspects of the waveforms and how these aspects are associated to each other, or to brain strains, we could gain a new appreciation of injury mechanisms. The eventual identification of impact patterns might allow us to differentiate between concussive and non-concussive hits, thus triggering sideline assessments and early identification of concussion. There may now be potential for highlighting dangerous patterns related to improper contact technique which could lead to recommendations for rule changes.

Conclusion

Head impact devices have allowed us to measure and study *in-vivo* head biomechanics in sports where concussions are frequent. Fifteen years of research has advanced our knowledge of concussion biomechanics but has also undermined our hopes of defining acceleration-based thresholds that could be used for identifying and preventing injuries. The first developed acceleration technology devices have presented important limitations restricting the possibilities for analysis. New developments and generalisation of innovative analytical approaches should allow sport scientists the ability to assess head impacts in new ways and further our understanding of concussion biomechanics.

Appendix H - Conference abstract: Validation of the linear acceleration measured by instrumented mouthguards for *in-vivo* head impact monitoring

This appendix was presented at the 2019 conference of the International Society of Biomechanics.

> Le Flao, E., Verma, K., Bourdet, N., King, D., Hume, P. A., Willinger, R., Borotkanics, R. J., Hamlin, M. J., & Takamori, S. (2019). Validation of the linear acceleration measured by instrumented mouthguards for *in-vivo* head impact monitoring. *International Society of Biomechanics,* Calgary, Canada.

Summary

Instrumented mouthguards are used to record *in-vivo* head kinematics in sport. The validation of the CSx mouthguard's capacity to measure linear acceleration was assessed using impact testing versus a reference single-axis accelerometer. The CSx mouthguard's linear acceleration demonstrated favourable validity; however, impact direction and impact magnitude resulted in differences between the mouthguard and the reference sensor.

Introduction

Concussions in sport can be caused by biomechanical forces following a direct impact to the head.[256] Instrumented mouthguards have been developed to measure skull motion during sports participation, providing insights into how the brain is affected by these forces. The challenge of associating concussion to skull motion is difficult because of measurement errors, and there is a need for valid and reliable measurement devices. This study aimed to assess the validity of a new instrumented mouthguard.

Methods

The CSx instrumented mouthguards (IMG) (CSx Systems Ltd, New Zealand) contained a triaxial accelerometer and triaxial gyroscope with sampling frequencies of 3.2 and 1 kHz, respectively. The validity of the linear acceleration of the IMG along each axis was assessed versus a reference single-axis accelerometer (REF) (Kistler SN C116114, Switzerland) using a sampling frequency of 10 kHz. IMG and REF were attached to an impactor with a 4.9 kg mass that was dropped using a guided rail experimental setup. Drop height ranged from 0.1 to 0.9 m to vary the range of impact magnitude. The impactor fell onto different padding materials to vary the range of impact durations. Three trials for each of the drop height and material combinations along X, Y and Z were repeated for a total of 54 impacts. Three variables were extracted from the output curves: peak (max value), duration (duration of the first section of the curve above 10 g) and impulse (area under the curve above 10 g). Linear regressions were performed between REF and IMG. Differences between the groups were analysed using Mann-Whitney's U test.

Results and Discussion

The measurements from IMG were consistent with the measurements from REF, as shown by good to very good coefficients of determination from linear regressions (Table H.1, Figure H.1). IMG recorded higher peak accelerations compared to REF, but this was statistically different for high magnitude impacts only (>130 g, 18-20%, $p < 0.05$). Differences across directions were present (Table H.1, Figure H.1). Impulse was greater for IMG on X (15%, $p = 0.017$). No other significant differences were found despite high mean differences between IMG and REF, especially along the X and Y directions (up to 19%). The absence of statistical differences might be explained by the variability between the impacts within a condition, IMG sampling frequency, the sensor's limitations, or the definition of the impact as per the 10 g-threshold.

Table H.1 Slopes and coefficients of determination from regressions between IMG and REF (all $p < 0.001$), all conditions combined.

	Peak (g)		Duration (ms)		Impulse (g.ms)	
	Slope	R2	Slope	R2	Slope	R2
X	1.38	0.99	1.31	0.86	1.23	0.92
Y	1.16	0.96	0.90	0.46	1.02	0.69
Z	1.07	0.99	1.04	0.92	1.10	0.98

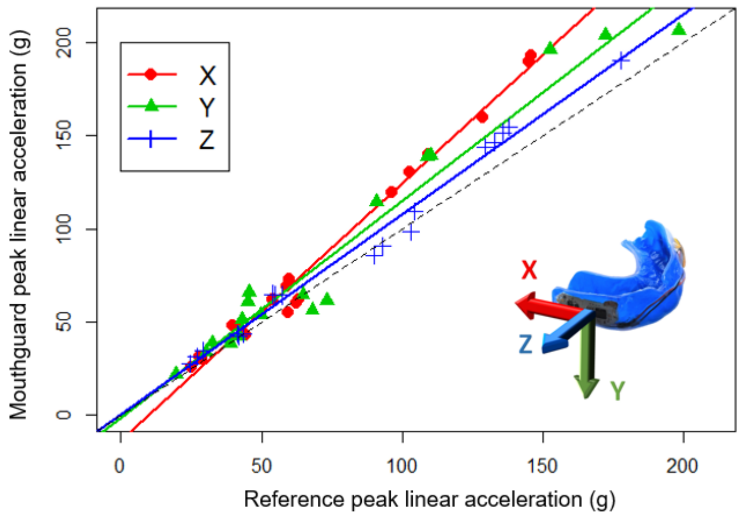


Figure H.1 Comparison of peak linear acceleration for IMG and REF along each axis. The dotted line represents IMG = REF.

Conclusions

Good consistency was achieved between the mouthguard and the reference under linear impacts along X and Z. This means that applying a correction factor to the mouthguard's measurements would be adequate. Differences were observed across axes and for high magnitude impacts. Further analyses, e.g. waveforms comparisons and mouthguard reliability, are ongoing.

Acknowledgments

The authors thank CSx for providing the mouthguards.

Appendix I - Assessing head/neck dynamic response to head perturbation: A systematic review.

This appendix presents a systematic review that was conducted prior to a change in scope for the PhD. The review has been published in *Sports Medicine*:

> Le Flao, E., Brughelli, M., Hume, P. A., & King, D. (2018). Assessing head/neck dynamic response to head perturbation: A systematic review. *Sports Medicine*, 48(11), 2641-2658. https://doi.org/10.1007/s40279-018-0984-3

Introduction

In collision sports such as rugby and American Football, participants engage in contacts such as tackles and collisions, exposing them to multiple impacts and stresses[131, 411] that can result in injuries to the head and neck.[108, 325] Cervical spine injuries and concussions are reportedly the most common injury types recorded in collision sports.[80, 382] Participation in collision sports and the accumulation of impacts to the head and neck can result in long-term impairments such as chronic neck pain,[35, 392] pathological changes in spinal morphology,[35, 392] neurocognitive deficits and psychological complications.[171, 240] These complications can affect players at all levels of participation.[171, 340] Therefore, the development of injury prevention strategies is crucial to assist with the management and prevention of head and neck injuries.

Head and neck injuries often occur simultaneously[75, 175, 262] and typically result from two types of events:[104, 202] (1) A "direct blow to the head, face, neck", where the neck is placed in tension by the head;[105, 413] or (2) A "direct blow on the body with an ''impulsive'' force transmitted to the head", where the neck has to prevent extreme head motion (i.e. whiplash-like situation).[256] The head's motion, in particular linear and rotational accelerations, is proposed to have a direct link to the risks of concussion and of microstructural and functional changes to the brain.[142, 239] As a result, accelerations have been studied *in-vivo* in many different sports.[115, 209, 281]

The cervical musculature contributes approximately 80% of the overall stability of the head/neck segment[305] and therefore may have an important protective role in injury reduction.[36, 82, 119, 218] The cervical musculature's reflex and voluntary contractions when the head is submitted to an impact are thought to protect against excessive movement, absorb energy of impacts, and reduce post-impact kinematic responses by changing the head/neck segment's stiffness and viscosity.[332, 371]

Previous studies in the sporting[241, 355] and automotive environments[217, 421] have assessed a variety of variables to better understand the musculoskeletal behaviour of the head and neck when the head is submitted to low magnitude impacts. Variables included displacement and acceleration of the head,[119, 217] stiffness,[371] and neck muscle activity.[217, 355] Co-variate effects of neck strength or awareness ('bracing for impact') on these variables have also been investigated[217, 241] as have the association between these parameters and concussion risks or head impact magnitudes.[82, 268]

However, there is limited evidence to suggest an association between cervical musculature capacities and head or neck injury risk reduction.[145, 256, 355] A better understanding of cervical musculature capacities, and of the effects of variables such as neck strength, is required to develop preventive strategies and reduce injuries and/or their consequences for athletes.

The aims of this review were therefore to: (a) compare and contrast methodologies that have utilised a mechanical perturbation to the head to assess head/neck dynamic responses of living human participants; (b) report on magnitude, validity and reliability of the methodologies; and (b) describe co-variates that may influence head/neck response.

Methods

The systematic review was registered with the International Prospective Register of Systematic Reviews (PROSPERO) on 7th December 2016 and was last updated on 27th February 2017 (registration number: CRD42016051057). Guidelines for the reporting of systematic reviews (PRISMA: Preferred Reporting Items for Systematic Reviews and Meta-Analyses)[279] and observational studies (STROBE: STrengthening the Reporting of OBservational studies in Epidemiology)[404] were followed. The PRISMA and STROBE guidelines contain checklists that were utilised for the conducting and reviewing of the included studies.

Search strategy and eligibility criteria

The title search field was alternately filled with text words arranged into combinations of [neck OR head OR cervical OR spine] AND [perturbation OR whiplash-like OR impact OR startle OR impuls* OR stability OR reflex OR stiffness]. The search strategy limited database results to academic journals, reviews, dissertations, and conference papers. The systematic search of the databases yielded 24,616 articles (Figure I.1) available online up to July 2018 through PubMed (N = 4,525), Web of Science Core collection (N = 6,529), SPORTDiscus with full-text (EBSCO) (1992-2016) (N = 401), Science Direct (N = 2,062), Scopus (N = 5,770), MEDLINE (OvidSP) (1946 to present) (N = 3,892) and CINAHL (EBSCO) (1997-2016) (N = 1,437). A comprehensive search of included articles, review of reference lists, and citation tracking on Google Scholar were utilised to identify additional relevant articles (N = 32). Duplicates were excluded at different stages of the screening process resulting in 8,974 references being retained.

All publications identified were initially screened by publication title and abstract to identify eligibility. There were no restrictions by study design or type of setting; reliability studies and studies with limited number of participants were included. Articles were included if: they were published in English of French; full text was available describing the methodology and device utilised; the intervention consisted of applying a direct perturbation to the head; a quantitative assessment of the head/neck response to the perturbation was provided. Studies were excluded if the intervention applied a perturbation to the body, or if the population of interest was not living adult humans.

Assessment of publication quality

All studies that met the inclusion criteria were assessed for quality by two authors on the basis of the STROBE checklist.[400, 404] For this review, quality was described as confidence that the study design, conduct and analysis minimized bias in estimation of the outcome measures. Initial agreement between the two authors was strong (Pearson's $r = 0.97$) and all disagreements were discussed until consensus was reached.

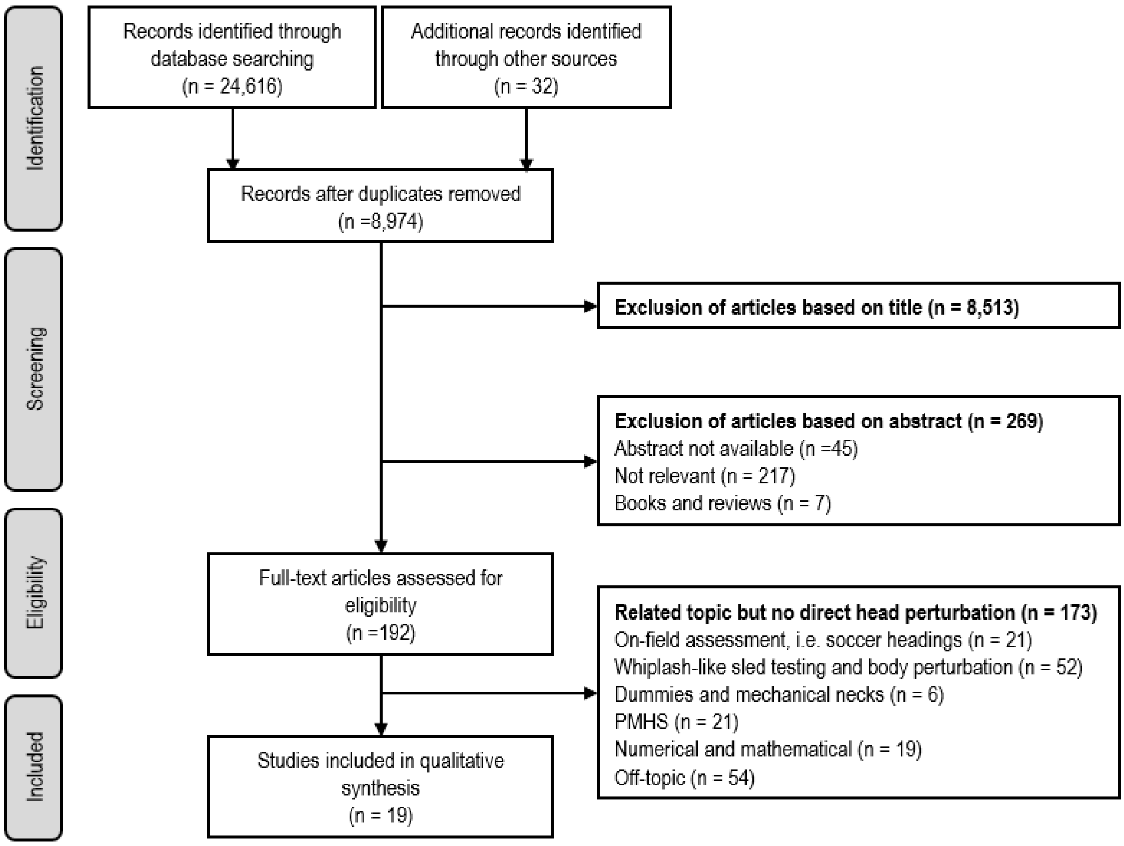


Figure I.1 Flow of identification, screening, eligibility and inclusion for the literature review of head/neck dynamic response to head perturbation. PMHS = Post-mortem human subjects.

Data extraction and treatment

The type of perturbation application, population and associated co-variates, and the conditions of perturbation were extracted from the articles. Two co-authors reviewed and came to a consensus on any extracted data requiring clarification. Methodological characteristics are summarised in Table I.1. Table I.2 presents the synthesis of the effects of co-variates on head/neck dynamic response. The results are written as a narrative,[317] taking care in reporting to minimise bias by ensuring the study quality did not influence the objective analysis of the methods used.

Statistical analyses

Summary measures of head/neck dynamic response to perturbation were undertaken for all studies and variables describing head/neck response. However, due to the heterogeneity between studies and inconsistency in reporting methods, these measures were not reported in this review. For these reasons, and therefore the lack of adequate sample size, a meta-analysis was not able to be performed.

Results

Article selection and quality assessment

Inclusion and exclusion criteria were used to select articles based on title (N = 461) then abstract (N = 192). 19 articles (15 journal articles,[9, 84, 119, 128, 130, 177, 178, 223, 241, 318, 320, 332, 355, 371, 391] two conference papers[319, 401] and two Master's theses[102, 235]) were finally reviewed and data were extracted for analysis based on consensus by two authors. The Master's theses were included because they included an article manuscript[102] or presented details on the apparatus[235] used in subsequent research.[401]

All studies were cross-sectional experimental studies except for one cohort study.[355] Several items of the STROBE list were not applicable to experimental designs, and were therefore excluded from the analysis. Some items were also not applicable to specific studies (i.e. item #12b "(b) Describe any methods used to examine subgroups and interactions" when no subgroups were analysed). The scores are presented as percentages to account for varying numbers of items across studies. This information is available in Table I.3.

Quality scores based on STROBE criteria for the included studies presented a median score of 79%, and ranged from 25% to 95%. The introduction and discussion were generally well documented and gave high scores for most studies according to the STROBE criteria. While the abstracts usually provided an informed and balanced summary of the study methods and results, the titles lacked a clear indication of the study design. Improvements in the rigor of reporting results might explain the lowest scores (≤ 61%) for the studies published before 1997.[84, 128, 177, 178,

[332] One Master's thesis[235] and one conference abstract[401] published after 1997 also had low scores. Other studies published after 1997 (score ≥ 75%) had their STROBE score reduced because they did not explain the study size or report participant characteristics.

Methodologies

Methodological approaches utilising a mechanical perturbation to the head to assess head/neck dynamic responses (Table I.1) included load dropping (11 publications),[9, 84, 102, 119, 128, 130, 241, 332, 355, 371, 391] quick release (6 publications),[84, 177, 178, 318-320] direct impact to the head via a pendulum,[223] and a motorized impactor.[235, 401] One study[84] reported outcome measurements for both load dropping and quick release methods.

Table I.1 Methodologies that have used a mechanical perturbation to the head to assess head/neck dynamic responses.

Reference	Population	Perturbation characteristics: Directions	Conditions	Preload	Energy
Load dropping studies					
Foust [128]	Healthy adults. Females, N = 93. Males, N = 77 (45%). Young adult (18-24 yr.), middle-age (35-44 yr.), elderly (62-74 yr.)	Flexion, extension	Unanticipated	57 g	0.45 - 0.89 J (455 g from 10-20 cm depending on the participant)
Reid et al.[332]	Young adults. Females, N = 1. Males, N = 7 (87%). 18 - 23 yr.	Extension	Various levels of awareness, and various instructions	Weight of cable and landing surface, not given	0.98 - potentially[a] 24.5 J (0.5 - 2.5 kg from 20 - 100 cm)
Corna et al.[84]	"Normal" participants. Females, N = 3. Males, N = 7 (70%). 22 - 49 yr. (N = 7 "normal" participants among these)	Rotation	Unanticipated, during movement. Active resisting or passive motion	Weight of cable and landing surface, not given	< 0.1 J (Impulsive momenta: 0.236 - 0.432 $kg.m.s^{-1}$, 2 kg from < 3mm)
Mansell et al [241]	NCAA collegiate soccer players. Females, N = 19, 19.16 ±0.90 yr. Males, N = 17 (47%), 19.21 ±0.92 yr.	Flexion, extension (randomized)	Anticipated, unanticipated (in that sequence)	Weight of cable and landing surface, not given	1.47 J (1 kg from 15cm)
Tierney et al.[391]	Physically active population. Females, N = 20, 24.2 ±4.1 yr. Males, N = 20 (50%), 26.3 ±4.3 yr.	Flexion, extension (randomized)	Anticipated, unanticipated (in that sequence)	Weight of cable and landing surface, not given	1.47 J (1 kg from 15 cm)
Fukushima et al.[130]	Young adults. Males = 10 (100%), 24.0 ±5.2 yr.	Extension	Level of awareness not mentioned. Pull from the forehead or at maxilla level.	Weight of cable and landing surface, not given	11.77 J (3kg from 40 cm landing on a spring)
Simoneau et al.[371]	Healthy adults. Females, N = 3. Males, N = 4 (57%). 23.5 yr.	Flexion, extension (in that sequence)	Unanticipated	2.22, 4.44, 6.67, 8.89 N (randomized)	0.49 J (1 kg from 5 cm)
Eckner et al.[119]	Contact sport athletes (soccer, ice hockey, US football, martial arts, wrestlers, lacrosse). Females, N = 22, 15.0 ±4.4 yr.; Males, N = 24 (52%), 16.3 ±5.0 yr. 8 - 30 yr.	Flexion, extension, lateral flexion (L), rotation (R) (not randomized)	Unanticipated, anticipated with maximum isometric co-contraction (in that sequence)	Weight of cable and landing surface counter-balanced by an opposite spring	0.15 - 0.78 J (1 kg from 1.5 to 8 cm, depending on BW)
Schmidt et al.[355]	High school and collegiate football players. Males, N = 49 (100%), 18.55 ±1.15 yr.	Flexion, extension (randomized)	Anticipated, unanticipated (in that sequence)	1 %BW	3.21 - 3.97 J (2.5 %BW mass from 15 cm)
Debison-Larabie[102]	Ice hockey players, Varsity and competitive leagues. Females, N = 8, 20.60 ±1.30 yr.; Males, N = 8 (50%), 22.13 ±1.55 yr.	Flexion, extension, lateral flexion (L&R) (randomized)	Anticipated, unanticipated	Weight of cable and landing surface, not given	2.21 J (1.5 kg from 15 cm)
Alsalaheen et al.[9]	Recreationally active young adults. Females N = 10, Males N = 9 (47%). 22.5 ± 1.7 yr. (range 18-25)	Extension	Anticipated, unanticipated (order not reported)	Preload (0.91 kg), no preload (in that sequence)	Kinetic energy equalling 3% of the subject's body mass.

Table I.1 (continued)

Reference	Population	Directions	Conditions	Preload	Energy
		Perturbation characteristics			
Quick release studies					
Ito et al.[177]	"Normal" adults. Females N = 2. Males N = 8 (80%). 21 - 49 yr. LDP. Males N = 4 (100%). 29 - 34 yr.	Extension (supine)	Unanticipated	No preload	Not applicable
Corna et al.[84]	"Normal" participants. Females N = 3. Males N = 7 (70%). Range 22 - 49 yr. LDP. Females N = 2. Males N = 4 (67%). 35 - 73 yr. (N = 10 "normal" participants and 4 LDP among these)	Flexion, rotation	Unanticipated, during movement	Females: 2 kg Males: 3 kg	Not applicable
Ito et al.[178]	"Normal" adults. Females N = 3. Males N = 7 (70%). 23 - 49 yr. LDP. Females N = 2. Males N = 4 (67%). 29 - 68 yr.	Extension (supine)	Unanticipated, with passive motion or active resisting (in that sequence)	No preload	Not applicable
Portero et al.[319]	Healthy adults. Females N = 2. Males N = 8 (80%). 30.6 yr.	Flexion, extension (randomized)	Unanticipated release	6 submaximal isometric force levels (from 20 to 70 %MVC)	Not applicable
Portero et al.[318]	Healthy adults. Females N = 2. Males N = 11 (85%). 27.1 ± 3.2 yr.	Flexion, extension (randomized)	Unanticipated	8 submaximal isometric force levels (from 10 to 80 %MVC)	Not applicable
Portero et al.[320]	Healthy adults. Females N = 2. Males N = 6 (75%). 31.7 ± 2.6 yr.	Flexion	Unanticipated release	Submaximal isometric force levels (from 20 to 70 %MVC)	Not applicable
Direct impact studies					
Kuramochi et al.[223]	Healthy adults. Males = 9 (100%). 22.6 ±4.1 yr. (19-30)	Extension	Anticipated, unanticipated (randomized)	No preload, no weight	1.66 J (pendulum of 4kg on a 45cm, 25° string)
Lucas[235]	Females N = 1 29 yr. Males N = 1 (50%) 23 yr.	Flexion, extension (randomized)	Unanticipated	No preload, no weight	Not reported (maximum force transmitted to the head: 12.7 ± 1.2 N)
Vasavada et al.[401]	Females N = 5. Males N = 4 (44%) age not given.	Extension, 45° rotated extension	Unanticipated	No preload, no weight	Not reported

Values are mean ± SD. SD = Standard Deviation; L = Left; R = Right; LDP = Labyrinthine-defective patients; MVC = Maximal Voluntary Contraction; BW = Body Weight; NCAA = National Collegiate Athletic Association.

[a] 24.5 J is the potential maximal value if the maximum height and weight were used as the study did not provide details as to the combination of height and weight.

Load dropping

Eleven studies[9, 84, 102, 119, 128, 130, 241, 332, 355, 371, 391] involved the sudden dropping of a load to induce head perturbation (Figure I.2). This methodology was utilised with athlete populations,[102, 119, 241, 355, 391] to investigate head/neck dynamic responses in traffic accidents (i.e. whiplash injury research with healthy adults),[128, 130] or to examine human neck reflex mechanisms.[9, 84, 332, 371]

The load dropping perturbation occurred by the impact of a free-falling weight on a landing surface. The landing surface was connected to the head via a non-extensible cable and a head harness,[9, 102, 119, 241, 332, 355, 391] a headband or strap,[128, 130, 371] or a plate held between the teeth[84] (Figure I.2). The cable ran through a height-adjustable pulley to ensure the force was applied perpendicular to the head/neck segment. The weight was either manually released[355] or via an electro-magnet.[84, 102, 128, 371] Only two studies utilised a safety stop to limit the displacement of the head set at 2.5 cm[119] or 10 cm[130] from its initial position. All load dropping studies investigated participant's responses in a seated position. The participants were static before the perturbation application in the majority of studies while in one study[84] they were actively moving a weight.

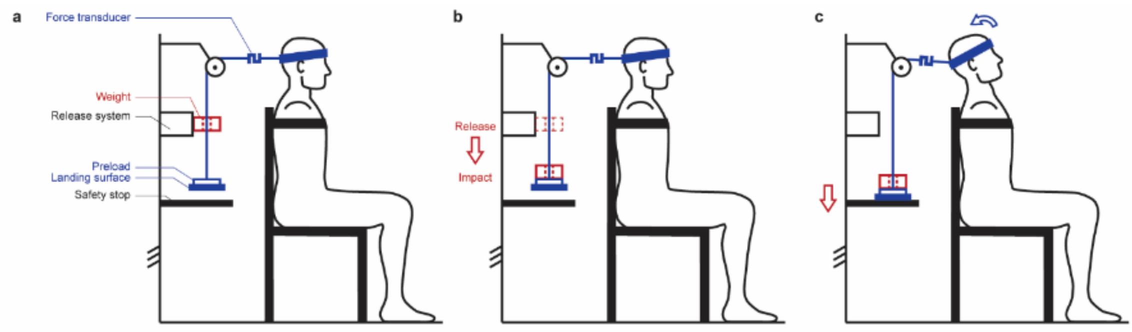


Figure I.2 Illustration of the load dropping method on neck extension. a) The participant sustains a generally static light load, with an optional preload mass; b) The weight is released and falls onto the landing surface; c) The impact created pulls on the participant's head.

Quick release

Six publications[84, 177, 178, 318-320] reported the use of quick release methods for assessment of head/neck segment dynamic properties. Testing conditions involved relaxed neck muscles,[177, 178] isometric sub-maximal contraction,[318-320] or a loaded dynamic movement.[84] In each case, the head was attached via a cable that was suddenly, and unexpectedly, released (Figure I.3, Figure I.4). Participants were healthy[318-320] or presented labyrinthine deficiency.[84, 177, 178]

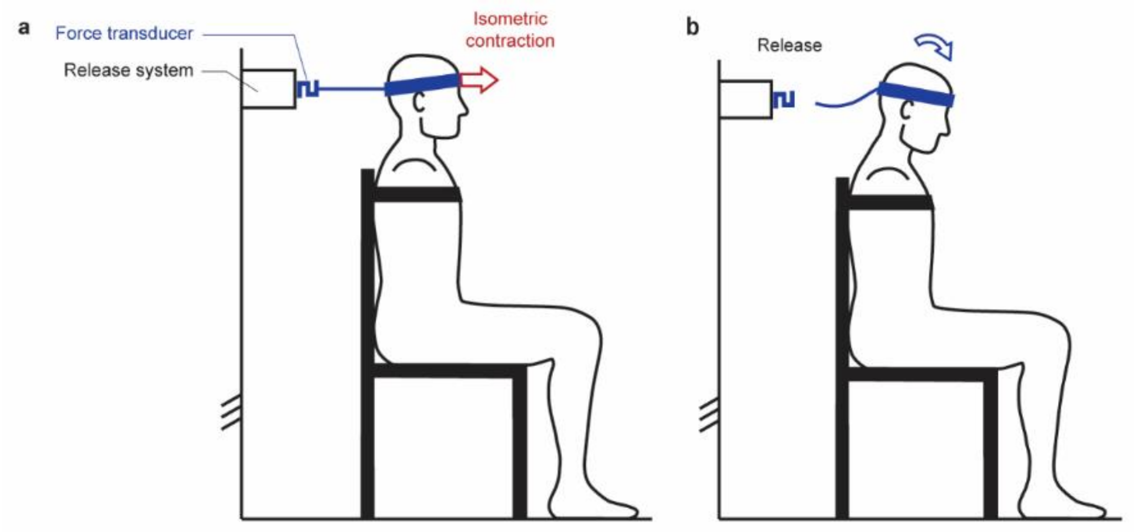


Figure I.3 Illustration of the quick release method in flexion as described by Portero et al.[318-320] a) Isometric contraction followed by b) the sudden and unexpected release of the cable leads to forward head motion

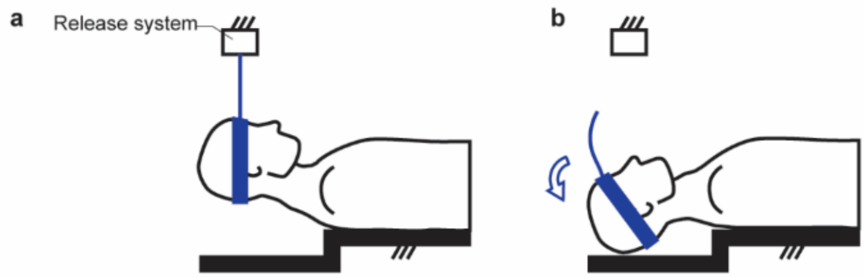


Figure I.4 Illustration of the quick release method as described by Ito et al.[177, 178] a) The participants lay supine with their head resting in a sling in a slightly flexed position. b) The sling is released and the head free falls to an extended position onto a cushioned landing surface

Direct contact

Three of the included studies involved a direct impact to the head, either with a ball on a pendulum to produce an impact to the forehead and force neck extension,[223] or with a linear motor that impulsed a hit at the vertex of a helmet worn by participants.[235, 401]

Factors to consider for assessment of head/neck dynamic responses

Torso restraint

To prevent upper body movement, and ensure that only neck muscles contributed to the movement, participants were reportedly restrained in a chair by a harness or strap at scapular, torso and/or lower trunk levels, in most but not all[9, 102] of the studies included. Two studies[9, 102] utilised a second investigator to check body position throughout the assessment. However, it has been reported[338] that neck isometric force production varies according to the location of the restraint on the torso. It is therefore recommended that a standardized thoracic restraint location at the level of the spine of the scapula be utilised to prevent upper body movement.

Instructions

As it is often the case in experiments involving human participation, variability can arise from the instructions given to participants.[246] In the articles included in this review, when participants were asked to react to the perturbation, the instructions utilised were reported to be "resist the load at its onset",[241] "as soon as",[391] or "once" they feel the tug,[9, 355] "maintain the head still",[371] or "right their head" and "resume tracking as quickly as possible".[84, 178] Analysis of the effects of different instructions on reaction times and computed neck stiffness[332] showed that muscle latency decreased from 90 ms when the participant was instructed to "resist as desired", and to 25 ms when the participant was instructed to "resist the tug as soon as possible". Neck stiffness reportedly doubled when the participant was instructed to "resist as much as possible" when compared to "resist as much as desired".[332]

Other studies[84, 177, 178, 223, 235, 401] assessed passive motion, where participants moved freely[401] with the perturbation without resisting. Passive motion, compared to active resisting, showed greater head displacement and peak velocity, as well as reduced muscular activity in healthy participants[84, 178] In the active quick release studies by Portero et al.,[318-320] the authors reportedly chose to study head movement in the first 15 to 30 ms, preceding reflex and voluntary muscle reactions, thus suppressing the need for instructions. The use of instructions was not reported in the other studies.[102, 119, 128, 130]

Researchers should be aware of the variability induced by instructions, and choose to study active resisting or passive motion depending upon their research question. In a sporting situation, it is more relevant to study active resisting as it is a human reflex to maintain the head in an upright position.[84, 178] To facilitate adequate comparison between studies, it is recommended that researchers systematically report the instructions given to their participants. Additional research is needed to determine the effects of the instructions on head kinematics and muscular response. This would give insights as to the best way to react to an unexpected head impact in order to minimize head/neck response.

Preloading

Included studies reported that the amount of preloading influences head/neck response.[9, 318-320, 371] Therefore, depending upon the load the participants must sustain before the onset of the perturbation, their consequent kinematics might vary. It is important for studies to report the preloading weight to facilitate comparison between studies. As head kinematics are also influenced by head inertia,[371] it is recommended that future experiments measure and report on the anthropometrics of the head and neck (e.g. head mass, head-neck segment length) and the weight of the headpiece worn by the participants.

Anticipation conditions

In a sporting context, head impacts and whiplash-like injuries can be sustained with or without the athlete being able to anticipate it.[264] Bracing for the impact allows an anticipatory co-contraction of the neck muscles to help reduce the consequences of the perturbation.[371] Therefore, depending upon the sport or situation being investigated, studying head/neck responses to both anticipated and unanticipated perturbations is recommended. All but one study[130] reported the participant's level of anticipation. All the quick release[84, 177, 178, 318-320] and four load dropping tests[84, 128, 371, 401] were performed with the participants systematically unaware of the onset of the perturbation. Other studies[9, 102, 119, 223, 241, 332, 391] explored the effects of anticipation on head and neck responses (the results are summarized in Table I.2 and in section Anticipation conditions in the studies), except for one study[355] where these were not reported or commented on.

The order of the conditions varied across the studies that tested both conditions. The anticipated trials were often performed first to let the participants accommodate safely to the impact.[223, 241, 355, 391] It is hypothesized that most experiments performing anticipated trials systematically before unanticipated trials chose to ensure participants' safety and comfort by doing so. This familiarization might however bias the unanticipated behaviour, as the perturbation level is not completely surprising.[391]

In an unanticipated trial, any noise or movement associated with the release of the perturbation mechanism can potentially influence the onset of muscular activity, be it voluntary or reflex.[84, 332] Therefore, noise and visible movement produced by the devices should be limited. Several protocols actively blocked visual and auditory cues, by closing the eyes,[102, 177, 178, 223] wearing blackened goggles,[241, 355, 391, 401] masking the perturbation device,[128] and/or noise cancelling devices.[223, 241, 355, 391, 401]

Analysis of multiple trials

Most studies included in this review reported that participants performed at least 3 trials for each condition and averaged the results of all trials. Some studies discarded the data from the first trial[9, 355] or the first three trials[235] of each series "*to eliminate a possible exaggerated neuromuscular startle response during the first exposure*". Indeed, in the first unanticipated trial, and especially when there have been no anticipated trials before, a 'startle' phenomenon can occur.[366] Siegmund et al.[366] defined this as a rapid protective response to an unexpected transient perturbation, the first exposure often evoking an exaggerated response. This reaction is typically a co-contraction of muscles, to stiffen the joints and protect against excessive movement. In simulated rear-end collisions, research has reported large reductions in neck muscles activity between the first and subsequent exposures.[295, 370] The reduction of the startle phenomenon with repeated exposures is called habituation.[370] Although it has never been studied in experiments with direct head perturbation, four of the reviewed studies[177, 223, 241, 332] mentioned the habituation phenomenon. Reid et al.[332] measured a drop of 50% in the stiffness between the first and second trial and attributed this to the participants realizing that the tug was not harmful and that they could relax. Ito et al.[177] noticed a greater muscular activity for the first trial, as well as a more pronounced eye blink. Mansell et al.[241] measured 40% differences in head accelerations and muscle onset latency between two test sessions separated by ~10 weeks. In contrast, Kuramochi et al.[223] suggested no habituation of the reflex response.

This raises several questions about averaging data from the first to the last trial, or excluding the first trial entirely. Despite a potential advantage of reducing variability, averaging across all trials would not be adequate if the data are obviously trending in a direction due to habituation. Excluding the first trial could however rule out useful information,[295, 366] especially in the study of unanticipated head impacts, where the analysis of the first trial only might be preferred. In conclusion, more research is needed to determine the evolution of head/neck response with trials sequence in head perturbation experiments. Researchers are also advised to associate their choice of methodology with their research question as all methodologies would provide different but useful information.

Body position, direction and location of force application

Most, but not all,[177, 178, 401] studies included in this review investigated head/neck response in a seated or supine position with the head in a neutral position. However, head impacts in sport rarely occur in such positions.[395] Studies have reported that body[354] and head position[144, 166] influence neck force production. For validity purposes, future research should capitalize on head impacts mechanisms and determine appropriate location and direction of impacts, as well as the body and head position, to replicate these conditions in the laboratory.

While most studies[9, 102, 128, 130, 332, 355, 371] applied the perturbation to the forehead for flexion or occiput for extension, some[241, 391, 401] chose to apply it to the vertex of the head, or had the participants holding a plate between their teeth.[84] However, Fukushima et al.[130] studied the effect of a horizontal force applied at the forehead or maxilla level on cervical vertebral movements. The results supported Reid's reported findings that head movement is a combination of translation and rotation.[130, 332] When the head is forced backwards by a horizontal force, the cervical spine adopts an S-shape, with flexion of its upper part, and extension of its lower part.[130] This phenomenon, termed 'cervical retraction',[130, 312] is even more pronounced when the perturbation is applied at maxilla level when compared to a forehead application. Therefore, it can be expected that the results would be different if applied at the vertex.

Mechanical loading

The energy of the impacts was not reported in studies utilising the load dropping[84, 102, 119, 128, 130, 241, 332, 355, 371, 391] and direct impact[223, 401] methods. One study[9] reported that the energy of the impact was chosen to equal 3% of the participant's body mass. The energy of the impacts could be calculated from the provided data when the height of fall and weight were provided, using the conservation of energy theorem. The resulting energies ranged from less than 0.1 J to potentially 24.5 J (see Table I.1). The majority (69%) of the studies[84, 102, 130, 223, 241, 332, 371, 391, 401] utilised the same energy for every participant, while other studies varied the weight[119, 355] or the fall height[9, 128] based upon participants' body weight. Only four studies[102, 223, 241, 391] presented comparable test conditions (non-normalised results for unanticipated perturbation in extension without preload in adult males without specific training). This included two studies with the same level of energy[241, 391] and one study[223] that did not report any common variable with the others. Therefore, the effect of impact energy on head/neck response could not be assessed. Because all load dropping and direct impact methods allow for control of the energy and the impact speed (by varying the weight, falling height, or the motor speed), evaluating head/neck responses over a range of impact magnitudes and speeds would be possible. While ensuring these magnitudes and speeds remain within safety limits, there is the potential to estimate the different responses over a range of various impact magnitudes.

The calculation of energy provides information about the impact, but does not account for the dampening properties of the apparatus.[119, 130] This includes shock absorption properties of the impacting components,[128, 130] tension of the cable, stretchiness and fitting of the headpiece.[119] Dampening of the perturbation modifies the force that is transmitted to the head, and most likely influences head kinematics.[130] It has been shown in rugby-related concussion research that hard-to-hard surface contacts (such as head-to-head, head-to-elbow) cause more head injuries than hard-to-soft contacts (such as head-to-upper body or head-to-lower leg),[395] making impact dampening a key factor for preventing injuries.

All load dropping studies measured the force transmitted to the head with a force transducer to determine perturbation onset time or neck stiffness.[149, 241, 355, 371, 391] However, this was only partially described in three studies[130, 241, 391] with the impact force reported to average 50[241, 391] to 200 N.[130] The characteristics of the force transmitted to the head, its magnitude, loading and unloading rates, and total duration, may have an influence on head kinematics and neck muscle reaction. It is also reasonable to accept that if participants are bracing for impact, and stiffening their neck instead of moving with the perturbation, the force will be higher and transmitted over a shorter duration. As a result, it is impossible to estimate the mechanical load that is actually applied to the head, limiting inter-study comparisons and analysis of the results. It is recommended that future studies provide information about dampening factors and real dynamic loading, indicating the peak force, the time to peak force, and the variability of these values.

Magnitude, reliability and variability of head/neck dynamic response variables

Magnitude

Eighteen different metrics have been utilised to report head/neck dynamic responses to perturbation, with the most frequent being neck muscle latency (9 studies, 18.6 to 88.0 ms),[9, 84, 102, 128, 177, 178, 241, 355, 391] neck stiffness (7 studies, 147.2 to 721.9 N.rad^{-1}, 14 to 1145.3 Nm.rad^{-1}),[241, 318-320, 355, 371, 391] and linear head acceleration (4 studies, 0.2 to 3.8 g).[84, 177, 178, 223]

Most, but not all,[9, 223, 319, 320] studies recorded head kinematics using 2D-[241, 355, 391] or 3D-motion capture,[102, 119, 401] accelerometers,[84, 128, 177, 178, 318, 332] angular velocity sensors,[371] or cineradiography.[130] The reported head kinematics variables varied greatly across studies, and included linear and/or angular peak accelerations, decelerations, velocities or displacements and time to peak acceleration (see Table I.2).

Sampling frequency of the included studies varied between 60 and 1500 Hz. However, Ito et al.[178] reported a time to peak acceleration as low as 9.9 ms. Because of this finding, the sampling frequency for head/neck response measurements should be at least 200 Hz, according to the Nyquist theorem.[406] Furthermore, it has been recommended[417] for real-time head impact measurements that the acceleration sampling frequency should be at least 500 Hz. For laboratory tests, where the impacts are not as demanding, the range of sampling frequency should be more than 200 Hz. However, it is recommended to consider higher frequencies to ensure adequate measurement of kinematic extrema and prevent ruling out important data, especially if efforts are made to limit dampening of the impact.

The choice of the variables was infrequently justified, but some kinematic metrics have been suggested to be associated with concussion,[148, 351, 423] whiplash injuries occurrence,[102] or brain tissue deformation.[119, 121] Linear and angular motions have different effects on brain injury mechanisms[202] and the movement caused by a direct perturbation to the head is a combination of translation and rotation.[130] Therefore, it is recommended that both linear and angular parameters are reported in future studies. When reporting linear parameters, it is also recommended that authors indicate the location of the point where they are measured, or project them to the estimated head's centre of gravity.[119]

Several composite injury metrics have also been proposed, associating several variables,[137, 147] but no study included in this review utilised these metrics to characterize head/neck response to a perturbation. It is important to acknowledge that despite peak linear and angular head accelerations being the most commonly reported variables in the studies of head impacts,[38, 149] there is no individual metric that is unequivocally accepted as being associated with brain injuries.[206, 306, 307]

Experiments often included surface electromyography (EMG) for measurement of muscle activity.[9, 84, 102, 128, 177, 178, 223, 235, 241, 332, 355, 391, 401] A few studies did not record muscular activity by EMG,[119, 318-320, 371] and one study[130] used EMG to control the participant's relaxed state before perturbation. Authors rarely reported peak[9, 241, 391] or mean[9, 102, 223, 241, 391] muscle activity, while all papers presented muscle onset latencies. Latencies were calculated utilising various methods, such as a change of magnitude by visual analysis[128, 332] or the precise calculation of activity threshold.[9, 355]

Each of the seven studies that investigated musculotendinous stiffness utilised either variations of Equation I.1,[241, 318-320, 332, 355, 391] or a more complex mathematical model to compute stiffness and viscosity.[371] Equation I.1 identifies that F is the force applied on the body and δ the displacement produced by the force.

$$k = \frac{F}{\delta} \qquad \textit{Equation I.1}$$

Despite reasonably similar protocols,[241, 355, 371, 391] comparison of stiffness results across studies is limited. Stiffness was reported using either force[241, 391] or torque[355, 371] measurements, and was normalized for some[355] but not all[241, 371, 391] studies by the participants' body weight.

A few studies normalized head kinematics[119] or neck stiffness[355] by the participants' body weight[119, 355] and height,[355] presuming there is a relationship between stiffness and body characteristics. This relationship is explained by the mathematical model utilised by Simoneau et al.[371] that describes stiffness as a generator of torque acting against gravity and perturbation forces.[371] From Newton's second law of motion, stiffness is expressed indirectly as a function of head and neck length and head mass, which can be estimated from body height and weight.[130, 241, 371, 391] Data normalization reduces inter-subject variability and it is therefore recommended that future studies normalise data, especially when comparing populations.

Reliability

Only two studies[318, 391] reported the reliability of the perturbation methodology providing test-retest results. Tierney[391] reported intraclass correlation coefficients (ICC) of 0.98 (95% CI 0.72 to 0.92) for head kinematics, 0.92 for neck muscles peak activity, 0.72 for muscle onset latency, and

0.96 for force measurement with the load dropping method. Portero et al.[318] reported an ICC of 0.81 to 0.96 and a standard error of measurement (SEM) of 0.9 to 2.2 Nm/deg for neck stiffness with the quick release method.

Validity

One study[355] compared the associations between head/neck response to perturbation and real head impacts measured by the Head Impact Telemetry System (HITS) over one football season. It was reported that increased stiffness of the flexor muscles was associated with reduced odds of sustaining moderate and severe head impacts (OR, 0.68; 95% CI, 0.48-0.96). Reactivity in the flexor muscles, i.e. a faster contraction of the cervical musculature, was also associated with decreased odds of sustaining severe head impacts (OR, 0.68; 95% CI, 0.49-0.95).[355] Being able to limit head angular displacement during perturbation testing has limited effects on the odds of sustaining higher magnitude impacts during sports participation (no odds ratios reached statistical significance).[355]

Load dropping and direct impact appear to be valid methods to replicate head impacts in sports. The main mechanisms reported for head injury in sports were direct hits to the head[104] either by another participant,[123, 196, 390] a moving object,[365] or the environment.[264] Pendulum[223] or motors[235, 401] were utilised to push the head, physically representing the head being hit. Furthermore, both methods involving pendulum or motor allowed the immediate release of the load after its application, permitting the head to move freely. For these reasons, these methods appear to re-create the most realistic sports-related head impacts. The load dropping method produces less realistic impacts as the head is being pulled. The resulting head motion might be similar, but it is unclear how the sensory stimulus of feeling a push or a pull influences neck muscular response.[177] Additionally, after the load has been dropped, it is maintained throughout the duration of the trial, and head movement is forced along the straight line formed between the head and the load, leading to less realistic conditions. However, the load dropping methods, when compared to the direct impact methods, allow easy measurement of the load that is applied to the head.

In comparison, the quick release methods do not represent an appropriate on-field situation as this method has been designed to isolate and study specific fundamental responses of the head/neck system.[84, 177, 178, 318-320] Specifically, the quick release methods utilised by Portero et al.[318-320] focused on characterizing passive musculotendinous stiffness during the 15 to 30 ms immediately after perturbation onset. This has the effect of inhibiting active dampening of the perturbation that occurs when the neck muscles contract reflexively or voluntarily. Sport scientists are interested in the whole head/neck response to head impact, and this involves muscular onset. However, the passive characteristics as described by Portero et al. might have an effect on head kinematics in the event of an unanticipated impact, when the muscles are not activated quickly enough to prevent from sudden movement.

Other quick release methods studies[84, 177, 178] have utilised this to characterize the effects of vestibular-collic and stretch-induced cervico-collic reflexes. The primary functions of these reflexes are to stabilize the head in space and on the trunk, respectively.[84] These studies identified that vestibular and stretch reflexes exhibited ~25 ms and ~65 ms latencies, respectively.[84, 177, 178] Therefore, the quick release methodologies[318-320] are not suitable for simulating sport-related head impacts, but can provide valuable information on passive characteristics and human reflex mechanisms.

The head was most often forced into extension or flexion. Each direction of motion activates muscles that may present different characteristics but inconsistency in the methodologies and reporting across the studies prevents any conclusions from being drawn. The choice of perturbation directions was justified in one study[355] based on the sport that was investigated. However, real-life head impacts rarely follow the anatomical planes,[310] and the choice of flexion- or extension-only perturbation warrants further discussion but is outside the scope of this review. The approach utilised by Schmidt et al.[355] in terms of composite metrics (the results summed across flexion and extension conditions) in assessing non-direction specific characteristics is proposed as a solution but also warrants further investigation.

There is limited human field-based evidence on the directional effects of concussion, although animal research and numerical modelling identify that direction of head motion influences brain response and injury risk.[143, 309, 423] Additional research is warranted to determine which neck muscles are involved in counter-acting injurious head impacts, and the direction in which to test in future perturbation studies. Additionally, most of the reviewed studies applied a linear load directed to the head's centre of gravity, generating mostly linear accelerations. Because it has been suggested[202] that rotational accelerations play a major role in concussion injury, future experiments focusing on rotational movement would give useful additional information.

Sports-related head impacts are usually characterized by linear and angular accelerations, and mean magnitudes for concussion have been estimated at 99 *g* and 5,777 rad.s^2.[38] Examination of the studies included in this review showed that the linear and angular accelerations reported did not exceed 4 g[84, 119, 177, 178, 223] and 42 rad.s^2,[102, 241, 391] or 4.2% and 0.7% of the estimated mean concussive accelerations, respectively. Also, 4 g is less than half the common threshold of 10 g, under which impacts are considered to be non-contact events and are excluded from analysis.[206] It is unknown if the results from the included studies hold true for greater magnitudes. More work is needed to determine if the laboratory tests are valid when compared with real head impact characteristics, not only in terms of acceleration magnitudes, but also of duration and loading rate. To the authors knowledge, neuromuscular characteristics, impact forces and stiffness have never been measured *in-vivo* and would provide researchers with useful information to help determine if laboratory experiments are realistic. Finally, despite peak linear acceleration being commonly reported in the head impact literature,[38] there is currently no consensus that this is the most appropriate variable to describe head/neck responses to real-life head impact with regards to concussion risk.[148, 214] It can be hypothesised that the lack of a validated variable led to that level of discrepancy in the reporting of head/neck responses. Well-designed prospective studies are warranted to investigate head/neck dynamic responses to real-life head impacts as risk factors for concussion.

Co-variates to consider in protocols and analyses of head/neck response

In all the perturbation investigations included in this review, participant populations varied in terms of sex (44% to 100% males), activity level (competitive athletes to non-sporting participants), age (8 to 74 years) and conditions of perturbation application such as anticipation and preloading (see Table I.1). The effects of these co-variates are reported in Table 2 for the 13 studies that reported information on co-variates. Studies not shown in Table I.2[84, 130, 177, 235, 332, 401] did not report any analysis of the co-variate effects.

Table I.2 Effects of co-variates on head/neck dynamic response to perturbation in 13 studies that reported co-variates such as sex, age, anthropometrics, anticipation, preloading, peak or rate of force/torque.

Co-variate; Study and method	Effects of co-variate
Sex	
Debison-Larabie[102] Load dropping	Females have: ↓ neck volume, ↑ HC/NC ratio, ↑ ang acc (Lflex, Flex), ↑ latencies (opposition SCM, SPN, SCL in Lflex, opposition SCM, SCL in Ext), ↓ latencies (all muscles in Flex), ↑ muscular activity (SPN, SCL-R in Flex, SCM, SCL in Ext), ↓ muscular activity (SCM, SCL-L in Flex), ↑ muscular activity in the reflex time period
Eckner et al.[119] Load dropping	No effect on peak lin and ang vel
Foust[128] Load dropping	Females have: ↓ latencies, ↑ head deceleration, ↓ time to peak deceleration
Mansell et al.[241] Load dropping	No sex differences in kinematics, EMG or stiffness.
Tierney et al.[391] Load dropping	Females have ↑ ang acc, ↑ ang disp, ↑ muscle activity (peak and area), ↓ latencies (SCM, Trap in Ext), ↓ stiffness
Age	
Eckner et al.[119] Load dropping	Effect of age on peak lin and ang vel ($P < 0.001$)
Foust[128] Load dropping	Elderly age group has: ↑ muscle latencies, ↑ time to peak deceleration
Ito et al.[178] Quick release	No correlation between age and muscle latencies
Anthropometrics	
Debison-Larabie[102] Load dropping	Weak to no relationship between HC/NC, neck volume or TNV and head ang acc
Eckner et al.[119] Load dropping	↑ CSA (SCM) = ↓ lin and ang vel in extension, ↑ NC = ↓ lin and ang vel for all directions
Foust[128] Load dropping	No effect of stature
Schmidt et al.[355] Load dropping	All players, larger SCM and SSC = ↑ odds[a]
Alsalaheen et al.[9] Load dropping	No association between NC or SCM CSA and neuromuscular response.
Isometric peak force/torque	
Eckner et al.[119] Load dropping	↓ lin vel for Ext, Flex ($P < 0.01$, $0.42 < R^2 < 0.63$); ↓ ang vel for Ext, Flex, Rot ($P < 0.01$, $0.43 < R^2 < 0.66$)
Schmidt et al.[355] Load dropping	All players: equal odds[a] between high and low performers (lin acc and HITSP). For linemen, stronger Lflex and Comp: ↑ odds[a]
Alsalaheen et al.[9] Load dropping	No association between peak force (Flex) and neuromuscular response.
Rate of force/torque development	
Eckner et al.[119] Load dropping	↓ lin vel for Ext, Flex ($P < 0.05$); ↓ ang vel for Ext, Flex, Rot ($P < 0.05$)
Schmidt et al.[355] Load dropping	All players, higher Ext RTD: ↑ odds of sustaining severe impacts. For skill players, higher Flx, Ext, Lflex, Comp RTD: ↑ odds[a]
Alsalaheen et al.[9] Load dropping	No association between RFD (Flex) and neuromuscular response.
Anticipated compared to unanticipated	
Debison-Larabie[102] Load dropping	↓ muscular activity in Lflex (1.4%)
Eckner et al.[119] Load dropping	↓ head lin (12.3%) and ang (9.7%) vel across all directions ($P < 0.001$)
Mansell et al.[241] Load dropping	↓ ang displacement (Ext: 23%, Flex: 25%), ↓ SCM peak activity (18%)
Tierney et al.[391] Load dropping	Males only: ↓ ang acc (25%)
Alsalaheen et al.[9] Load dropping	↑ pre-impact muscular activity (SCM), ↓ onset latency (SCM). No effect on average and peak muscular activity and time to peak muscular activity.
Ito et al.[178] Quick release	↓ peak head lin vel
Kuramochi et al.[223] Direct impact	↓ muscular activity (SCM, Ext)
Preloading	
Portero et al.[318] Quick release	↑ head ang disp ($0.94 < R^2 < 0.99$)
Portero et al.[319] Quick release	↑ stiffness ($0.45 < R^2 < 0.68$)
Portero et al.[320] Quick release	↑ stiffness ($R^2 = 0.74$)
Simoneau et al.[371] Load dropping	↓ peak ang vel, (-18.2% to -19.9%), ↓ time to peak ang vel (-15%), ↑ stiffness (+29.8% to +36.3%), ↑ viscosity (+27.4% to +31.0%)
Alsalaheen et al.[9] Load dropping	↑ pre-impact muscular activity (SCM), ↑ average post-impact muscular activity, ↓ onset latency (SCM). No effect on peak muscular activity and time to peak muscular activity.

a = odds of sustaining a moderate or severe head impacts compared to mild head impacts.[355] HC/NC = head circumference/ neck circumference; TNV = Total Neck Volume; CSA = Cross-sectional area; NC = Neck circumference; EMG = electromyography; ang = angular; liN = linear; acc = acceleration; vel = velocity; disp = displacement; RTD = rate of torque development; SCM = Sternocleidomastoid; SSC = Semispinalis capitis; SPN = Splenius capitis; SCL = Scalene; SCL-R = right scalene; SCL-L = left scalene; Trap = Trapezius; Flex = Flexion; Ext = Extension; Lflex = Lateral Flexion; Rot = Axial rotation; HIT_{SP} = Head impact telemetry severity profile; Comp = Composite.

Sex

Some studies[102, 128, 391] reported males had slower reflex times when compared with females. It was reported that female muscles were able to start contracting more promptly, although this is not corroborated for all directions of perturbation.[102] Females also seem to be able to decelerate their head motion faster and stronger than males do.[128] However, it has been reported[102, 391] that despite contracting earlier, or at the same time, and using an equal or greater proportion of their muscular abilities, females exhibited more head angular acceleration and displacement and similar velocities.[119] This is consistent with Newton's law of acceleration as for a given force application, less head mass correlates with greater acceleration, and women displayed less head mass when compared to men.[391] Reduced neck strength and stiffness were also reported for females, and may be related to their smaller amount of muscular tissue.[102, 391] Mansell et al.[241] did not observe any sex-related differences, and attributed this to the level of physical activity of the participants. The participants were soccer players, trained in heading the ball, and might have had greater neck neuromuscular capacities when compared with a less active population. The authors[241] suggested this would have reduced the differences between the sexes.

To date, there is no consensus on sex as a risk factor for concussion[256] but there is evidence that sex differences do exist in the outcomes of concussion.[109] Until more research is undertaken to identify if neck dynamic properties are risk factors for concussion, it is difficult to use sex differences for practical preventive applications, but it should be considered as a modifying factor for head/neck dynamic response to perturbation.

Age

There appears to be an age effect on neck strength and head/neck response to perturbation. Younger (high school or younger)[119] and elderly[128] participants exhibited less neck strength, higher linear and angular velocities or increased muscle latencies when compared with young and middle-aged adults. There does not seem to be an age effect for participants 18 to 50 years old.[128, 178] There is a potential interaction between sex and age, with adult males' strength capacities peaking in the middle-age category, and females' in the young adult category.[128]

Anthropometrics, neck strength and force development

Stature did not influence head/neck dynamic response.[128] Head and neck dimensions, including muscle volume, might have a slight effect on head kinematics[102, 119] but not on neuromuscular response.[9] However, it seems that larger cervical muscles decrease perturbation kinematics,[119] but increase the odds of sustaining high magnitude head impacts during football play.[355]

Six studies[9, 119, 128, 241, 355, 391] assessed isometric neck strength or rate of force development (RFD). These parameters were tested for their effects on the dynamic response to the perturbation. There were significant correlations (r = 0.417 to 0.657) between neck strength and linear and angular head velocities in most directions of perturbation.[119] Other studies[128, 241, 391] have not tested the direct association between neck strength and head kinematics, but from the limited data available for quantitative analysis (not reported here) there were contradicting trends. In the study by Tierney et al.,[391] the results suggest potential relationships between isometric neck strength and peak angular acceleration ($r^2 = 0.90$), displacement ($r^2 = 0.99$) and stiffness ($r^2 = 0.70$), but these trends were not seen in the other studies.[128, 241] Corroborating the absence of relationship, in an intervention study of 36 soccer players,[241] a significant increase (15% increase, $p < 0.001$) in isometric neck strength achieved by resistance training did not alter the participant's reactive muscle activity or head kinematics. This is further supported by the absence of an association between isometric peak force and neuromuscular response.[9] Furthermore, in the only study[355] investigating real-life head impacts, football players with significantly ($p < 0.001$) stronger, larger necks had equal odds of sustaining higher magnitude head impacts during games when compared to players with weaker, thinner necks (odds ratios ranging from 0.88 to 1.65).[355]

While there does not seem to be an association between RFD and neck neuromuscular response,[9] Eckner et al.[119] identified that a higher RFD reduced head kinematics in several directions. This is further illustrated by two studies utilising the same perturbation protocol, that showed that trained soccer players[241] exhibited less head angular acceleration than physically active participants.[391] This suggests that specific soccer training such as heading the ball might influence head/neck dynamic responses by enhancing neuromuscular control. Because of methodological and reporting differences, it is not possible to verify if this holds true with football[355] or ice hockey players.[102] Future research is warranted to investigate how various neck strengthening and conditioning exercises could improve stiffness and muscle reactivity, which have been shown to be associated with a reduction in the magnitude of real-life head impacts.[355]

In conclusion, it is unclear if cervical strength influences head/neck dynamic responses and real-game accelerations and if these influence the incidence of concussions. Several studies[145, 241, 355] have recommended neuromuscular training, such as plyometrics, to stimulate short-latency force production, enhancing motor control mechanisms and joint dynamic stabilization. Future research is warranted to investigate the effects of muscular short-latency force production and of fatigue on neck muscle capacities, and how these impact on the response to perturbations and the risk of concussions.

Anticipation conditions in the studies

The effects of anticipation on head motion are contradictory, with studies reporting similar[102, 223, 241] or decreased[119, 241, 391] head acceleration, velocity, and displacement when compared with unanticipated perturbations. Neck stiffness showed no significant ($p>0.05$) difference between anticipation conditions.[241, 391] Neck muscle activity was reduced when participants braced for the impact in most[102, 223, 241] but not all[9, 391] studies. Muscle onset latency was reduced[9] as well as more efficient[332] (i.e. meaning the participant postponed muscle activity until immediately prior to the onset of the perturbation) when the participants knew of the perturbation onset.[332] Anticipation ("bracing for the impact") reduces head/neck response but the relevance of studying unanticipated response depends on the sport and the situation being investigated.

Preloading of perturbation

In the load dropping studies, participants had to sustain a certain load before the onset of the perturbation, which consisted of the weight of the cable and of the landing surface.[9, 84, 102, 128, 130, 241, 332, 391] A spring was utilised in one study[119] to counter-balance the initial static force applied to the participant's head, but the spring characteristics were not reported. Preloading weights were also utilised to simulate active bracing by increasing neck muscle contraction[9, 355, 371] and study its effects on head kinematics, neck viscoelastic properties[371] and neuromuscular response.[9] It has been reported[371] that pre-perturbation loading is positively associated with increased muscular activity,[9, 371] neck stiffness and viscosity,[371] and decreased head kinematics[371] and muscle onset latency.[9] These results are further supported by quick release experiments[319, 320] where neck stiffness was dependent upon the torque applied before release.

Conclusions

Methods utilised in testing neck/head dynamic responses to perturbation included load dropping, quick release, and direct contact with the head. Based on validity, ease of use and of configuration, the best method for the simulation and study of sport-related head impacts appears to be the direct impact method via pendulum. However, this method does not allow the measurement of the force transmitted to the head during the perturbation, making the calculation of neck stiffness challenging. For this reason, the load dropping method is recommended.

The magnitude of the perturbation should be kept under 4 J for participants' safety if no dampening system is used, and future research should investigate how the magnitude influences head/neck response.

There was inconsistency in the variables chosen to describe head/neck dynamic responses to a perturbation. Studies reviewed reported different head kinematics, neck neuromuscular variables, neck stiffness and viscosity. Due to discrepancies in experimental protocols and reporting processes, it was not possible to summarize the results quantitatively. As a result, a narrative analysis was performed to summarize the effects of co-variates.

When undertaking research, it is recommended that age should be a consideration, especially if the population is outside of the 18-50-yr. age range as youth and elderly populations present notably weaker head/neck responses. Other modifying factors include sex, neck force production and neuromuscular control. It is also recommended that future research reflects on body and head position, direction of perturbation, and anticipation conditions with regards to the characteristics of the sport under investigation.

Dynamic responses such as neck stiffness and muscle onset latency have been identified as possible risk factors associated with head injuries or high-magnitude head accelerations, but this has only been shown in one study to our knowledge.[355] In the case of a foreseeable impact, the capacity of a player to brace for the impact seems to be key, as shown by several studies reporting reduced head kinematics in association with bracing by anticipation and/or preloading. As the investigators also measured an increase of pre-impact muscular activity in those cases, it is hypothesized that increased muscle contraction increases neck stiffness, leading to reduced head kinematics. The capacity to quickly produce a great amount of strength at the right moment to absorb the impact might be the key characteristic involved, but the evidence is scarce and conflicting.[9, 119, 355] Further research is warranted to identify head/neck dynamic response variables related to injury risk.

In conclusion, each methodology can provide useful information on the head/neck dynamic response. However, the validity and relevance of these methodologies when compared to *in-vivo* impact measurement still needs to be addressed. Reports on head/neck response should include neck muscle latency (ms), neck stiffness ($N.rad^{-1}$ or $Nm.rad^{-1}$) and linear (g) and rotational ($rad.s^{-2}$) head accelerations given the suggested validity of these metrics with respect to concussion risks. Modifying factors for head/neck dynamic response that need to be considered are anticipation and participants' age, sex, and sports participation.

Acknowledgements

The authors thank the Auckland University of Technology for the AUT SRIF RCRG Neck Strength Project grant 2016.

Compliance with Ethical Standards

Funding

This systematic review formed part of the AUT SRIF RCRG Neck Strength Project grant 2016. This funding supported development of the research strategy, data extraction and analyses by the

primary investigators. The funder had no input into the interpretation or publication of the study results.

Conflicts of Interests

Enora Le Flao, Matt Brughelli, Patria A. Hume and Doug King declare that they have no conflicts of interest relevant to the content of this review.

Author Contributions

According to the definition given by the International Committee of Medical Journal Editors (ICMJE), the authors listed above qualify for authorship based on making one or more substantial contributions to the intellectual content of the manuscript. Enora Le Flao and Matt Brughelli were responsible for the conception and design of the review, and the acquisition, analysis and interpretation of data; they also contributed to the drafting of the manuscript and critical revision. In addition, Matt Brughelli contributed to funding acquisition. Doug King contributed to the drafting of the manuscript and to critical revision. Patria Hume contributed to critical revision and funding acquisition.

Electronic Supplementary Materials

Table I.3 (next pages) STrengthening the Reporting of OBservational studies in Epidemiology (STROBE)350 assessment of reviewed publications.

Item		Recommendation	Reid et al.[332]	Vasavada et al.[401]	Ito et al.[177]	Ito et al.[178]	Corna et al.[84]	Lucas[235]	Foust[128]	Fukushima et al.[130]	Debison-Larabie[102]	Kuramochi et al.[223]	Alsalaheen et al.[9]	Portero et al.[319]	Portero et al.[320]	Eckner et al.[119]	Tierney et al.[391]	Portero et al.[318]	Simoneau et al.[371]	Schmidt et al.[355]	Mansell et al.[241]	Item scores
Title and abstract																						
Title and abstract	1	(a) Indicate the study's design with a commonly used term in the title or the abstract	0	0	0	0	0	0	0	0	0	0	0	0	0	0.5	0	0.5	0	0.5	0.5	21%
		(b) Provide in the abstract an informative and balanced summary of what was done and what was found	0	0	0.5	0.5	0.5	0.5	0.5	0.5	0.5	0	0.5	0.5	0.5	0.5	0.5	0.5	0.5	0.5	0.5	84%
Introduction																						
Background/ rationale	2	Explain the scientific background and rationale for the investigation being reported	1	1	1	1	1	1	1	1	1	1	1	1	1	1	1	1	1	1	1	100%
Objectives	3	State specific objectives, including any prespecified hypotheses	0	1	0	1	0	1	1	1	1	1	1	1	1	1	1	1	1	1	1	84%
Methods																						
Study design	4	Present key elements of study design early in the paper.	1	0	0	0	1	0	0	1	0	1	0	1	1	1	1	0	1	1	1	58%
Setting	5	Describe the setting, locations, and relevant dates, including periods of recruitment, exposure, follow-up, and data collection.	0	0	1	1	1	0	1	1	1	1	0	1	1	1	1	1	1	1	1	79%
Participants	6	Give the eligibility criteria, and the sources and methods of selection of participants. Describe methods of follow-up for cohort studies.	0	0	0	0	0	0	1	1	1	1	1	1	1	1	1	1	1	1	1	68%
Variables	7	Clearly define all outcomes, exposures, predictors, potential confounders, and effect modifiers. Give diagnostic criteria, if applicable.	1	0	0	0	0	0	1	1	0	0	1	1	1	1	1	1	1	1	1	63%
Data sources/ measurement	8	For each variable of interest, give sources of data and details of methods of assessment. Describe comparability of assessment methods if there is more than one group	0	1	0	0	0	1	1	1	1	1	1	1	1	1	1	1	1	1	1	79%
Bias	9	Describe any efforts to address potential sources of bias	0	1	0	0	0	1	0	1	1	1	1	1	1	1	1	1	1	1	1	74%
Study size	10	Explain how the study size was arrived at	0	0	0	0	0	0	0	0	0	0	0	0	0	0	0	0	0	0	1	5%
Quantitative variables	11	Explain how quantitative variables were handled in the analyses. If applicable, describe which groupings were chosen and why	0	0	0	0	0	1	0	1	1	1	1	1	1	1	1	1	1	1	1	68%
Statistical methods	12	(a) Describe all statistical methods, including those used to control for confounding	0	0.5	0	0	0	0.5	0	0	0.5	0.5	0.5	0.5	0.5	0.5	0.5	0.5	0.5	0.5	0.5	68%
		(b) Describe any methods used to examine subgroups and interactions (NA if no subgroups)	NA	NA	0	0	0	0.5	0	NA	0.5	0.5	0.5	NA	NA	0.5	0.5	NA	0.5	0.5	0.5	69%

Table I.3 (continued)

Item		Recommendation	Reid et al.[332]	Vasavada et al.[401]	Ito et al.[177]	Ito et al.[178]	Corna et al.[84]	Lucas[235]	Foust[128]	Fukushima et al.[130]	Debison-Larabie[102]	Kuramochi et al.[223]	Alsalaheen et al.[9]	Portero et al.[319]	Portero et al.[320]	Eckner et al.[119]	Tierney et al.[391]	Portero et al.[318]	Simoneau et al.[371]	Schmidt et al.[355]	Mansell et al.[241]	Item scores
Results																						
Participants	13	(a) Report numbers of individuals at each stage of study—eg numbers potentially eligible, examined for eligibility, confirmed eligible, included in the study, completing follow-up, and analysed	0	1	1	1	1	1	1	1	1	1	1	1	1	1	1	1	1	1	1	95%
Descriptive data	14	(a) Give characteristics of study participants (eg demographic, clinical, social) and information on exposures and potential confounders	0	0	0.5	0.5	0.5	0	0.5	0	0.5	0	0	0	0	0.5	0.5	0.5	0	0.33	0.5	51%
		(b) Indicate number of participants with missing data for each variable of interest	0	0.5	0	0	0	0.5	0.5	0.5	0	0.5	0	0	0	0	0.5	0	0	0.33	0.5	40%
		(c) Cohort study—Summarise follow-up time (eg, average and total amount)																		0.33		
Outcome data	15	Report numbers of outcome events or summary measures over time	0	0	1	1	1	1	1	0	1	1	1	1	1	1	1	1	1	1	1	84%
Main results	16	(a) Give unadjusted estimates and, if applicable, confounder-adjusted estimates and their precision (eg, 95% confidence interval). Make clear which confounders were adjusted for and why they were included	0	0	0	0	1	0	0	0	1	1	1	0	0	0	0	1	1	1	1	42%
Other analyses	17	Report other analyses done—eg analyses of subgroups and interactions, and sensitivity analyses (NA if no other analyses)	NA	NA	0	0	0	0	0	NA	1	NA	1	NA	1	1	1	1	1	1	1	64%

Table I.3 (continued)

Item		Recommendation	Reid et al.[332]	Vasavada et al.[401]	Ito et al.[177]	Ito et al.[178]	Corna et al.[84]	Lucas[235]	Foust[128]	Fukushima et al.[130]	Debison-Larabie[102]	Kuramochi et al.[223]	Alsalaheen et al.[9]	Portero et al.[319]	Portero et al.[320]	Eckner et al.[119]	Tierney et al.[391]	Portero et al.[318]	Simoneau et al.[371]	Schmidt et al.[355]	Mansell et al.[241]	Item scores
Discussion																						
Key results	18	Summarise key results with reference to study objectives	1	1	1	1	1	1	1	1	1	1	1	1	1	1	1	1	1	1	1	100%
Limitations	19	Discuss limitations of the study, taking into account sources of potential bias or imprecision. Discuss both direction and magnitude of any potential bias	0	0	1	1	1	1	1	1	1	1	1	1	1	1	1	1	1	1	1	89%
Interpretation	20	Give a cautious overall interpretation of results considering objectives, limitations, multiplicity of analyses, results from similar studies, and other relevant evidence	0	0	1	0	1	1	1	1	1	1	1	1	1	1	1	1	1	1	1	84%
Generalisability	21	Discuss the generalisability (external validity) of the study results	1	0	1	1	1	1	1	1	1	1	1	1	1	1	1	1	1	1	1	95%
Other information																						
Funding	22	Give the source of funding and the role of the funders for the present study and, if applicable, for the original study on which the present article is based	0	1	1	1	1	0	0	0	0	0	1	0	0	0	0	0	1	0	0	32%
Absolute score			5	8	10	10	12	13	13.5	15	17	16.5	17.5	16	17	18.5	18.5	18	19.5	20.0	21	
Relative score (accounting for not applicable items)			25%	40%	45%	45%	55%	59%	61%	75%	77%	79%	80%	80%	81%	84%	84%	86%	89%	91%	95%	