Sports Medicine
https://doi.org/10.1007/s40279-018-0984-3

SYSTEMATIC REVIEW

# Assessing Head/Neck Dynamic Response to Head Perturbation: A Systematic Review

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## Abstract

**Background** Head/neck dynamic response to perturbation has been proposed as a risk factor for sports-related concussion.
**Objectives** The aim of this systematic review was to compare methodologies utilised to assess head/neck dynamic response to perturbation, report on magnitude, validity and reliability of the response, and to describe modifying factors.
**Methods** A systematic search of databases resulted in 19 articles that met the inclusion and exclusion criteria.
**Results** Perturbation methods for head/neck dynamic response included load dropping, quick release and direct impact. Magnitudes of perturbation energy varied from 0.1 to 11.8 J. Head/neck response was reported as neck muscle latency (18.6–88.0 ms), neck stiffness (147.2–721.9 N/rad, 14–1145.3 Nm/rad) and head acceleration (0.2–3.8*g*). Reliability was only reported in two studies. Modifying factors for head/neck response included younger and older participants presenting increased responses, females showing better muscular reactivity but similar or increased head kinematics compared with males, and bracing for impact limiting muscular activity and head kinematics.
**Discussion** Substantial differences in experimental and reporting methodologies limited comparison of results. Methodological factors such as impact magnitude should be considered in future research.
**Conclusion** Each methodology provides valuable information but their validity for anticipated and unanticipated head impacts measured in vivo needs to be addressed. Reports on head/neck response should include measurement of transmitted force, neck muscle latency, head linear and rotational accelerations, and neck stiffness. Modifying factors of anticipation, participants' age, sex, and sport are to be considered for head/neck dynamic response.
**PROSPERO Registration Number** CRD42016051057 (last updated on 27 February 2017).

**Key Points**

Modifying factors for head/neck dynamic response that need to be considered in studies are anticipation ('bracing for impact') and participants' age, sex, and sports participation.

There was initial evidence for neck muscle reactivity and neck stiffness being associated with risks of sustaining high-magnitude head impacts.

## 1 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 [1, 2] that can result in injuries to the head and neck [3, 4]. Cervical

**Electronic supplementary material** The online version of this article (https://doi.org/10.1007/s40279-018-0984-3) contains supplementary material, which is available to authorized users.

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Published online: 21 September 2018

spine injuries and concussions are reportedly the most common injury types recorded in collision sports [5, 6]. 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 [7, 8], pathological changes in spinal morphology [7, 8], neurocognitive deficits and psychological complications [9, 10]. These complications can affect players at all levels of participation [9, 11]; 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 [12–14] and typically result from two types of events [15, 16]: (1) a 'direct blow to the head, face, neck', where the neck is placed in tension by the head [17, 18]; 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. a whiplash-like situation) [19]. The head's motion, in particular linear and rotational accelerations, is proposed to have a direct link to the risks of concussion and to microstructural and functional changes to the brain [20, 21]. As a result, accelerations have been studied in vivo in many different sports [22–24].

The cervical musculature contributes approximately 80% of the overall stability of the head/neck segment [25] and may therefore have an important protective role in injury reduction [26–29]. 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 [30, 31].

Previous studies in the sporting [32, 33] and automotive environments [34, 35] 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 [29, 34], stiffness [31], and neck muscle activity [33, 34]. Covariate effects of neck strength or awareness ('bracing for impact') on these variables have also been investigated [32, 34], as have the association between these parameters and concussion risks or head impact magnitudes [28, 36].

However, there is limited evidence to suggest an association between cervical musculature capacities and head or neck injury risk reduction [19, 33, 37]. A better understanding of cervical musculature capacities and 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 (1) compare and contrast methodologies that have utilised a mechanical perturbation to the head to assess head/neck dynamic responses of living human participants; (2) report on magnitude, validity and reliability of the methodologies; and (3) describe covariates that may influence head/neck response.

## 2 Methods

This systematic review was registered with the International Prospective Register of Systematic Reviews (PROSPERO) on 7 December 2016 and was last updated on 27 February 2017 (registration number CRD42016051057). Guidelines for the reporting of systematic reviews (PRISMA: Preferred Reporting Items for Systematic Reviews and Meta-Analyses [38]) and observational studies (STROBE: STrengthening the Reporting of OBservational studies in Epidemiology [39]) were followed. The PRISMA and STROBE guidelines contain checklists that were utilised for the conducting and reviewing of the included studies.

### 2.1 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 (see Fig. 1) available online to July 2018 through PubMed ($N=4525$), Web of Science Core collection ($N=6{,}529$), SPORTDiscus with full-text [EBSCO; 1992–2016] ($N=401$), ScienceDirect ($N=2062$), Scopus ($N=5770$), MEDLINE [OvidSP; 1946–present] ($N=3892$) and CINAHL [EBSCO; 1997–2016] ($N=1437$). 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 8974 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 a limited number of participants were included. Articles were included if they were published in English or French; full text was available describing the methodology and device utilised; the intervention consisted of applying a direct perturbation to the head; and 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.

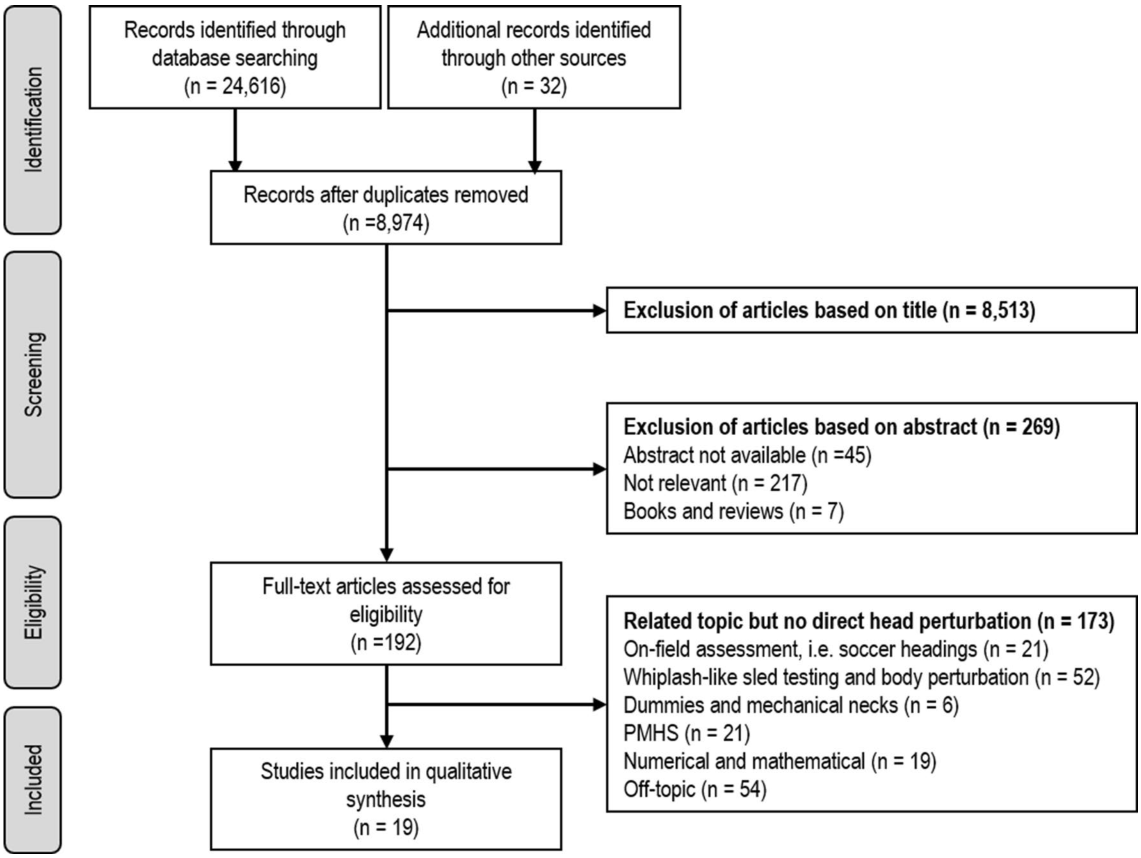


**Fig. 1** Flow of identification, screening, eligibility and inclusion for the literature review of head/neck dynamic response to head perturbation. *PMHS* post-mortem human subjects

### 2.2 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 [39, 40]. 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.

### 2.3 Data Extraction and Treatment

The type of perturbation application, population and associated covariates, and the conditions of perturbation, were extracted from the articles. Two co-authors reviewed the data and came to a consensus on any extracted data requiring clarification. Methodological characteristics are summarised in Table 1, while Table 2 presents the synthesis of the effects of covariates on head/neck dynamic response. The results are written as a narrative [41], taking care in reporting to minimise bias by ensuring the study quality did not influence the objective analysis of the methods used.

### 2.4 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.

## 3 Results

### 3.1 Article Selection and Quality Assessment

Inclusion and exclusion criteria were used to select articles based on title ($N = 461$) and then abstract ($N = 192$). Overall, 19 articles (15 journal articles [29–33, 42–51], two conference papers [52, 53] and two Master's theses [54, 55]) 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 [54]

**Table 1** Methodologies that have used a mechanical perturbation to the head to assess head/neck dynamic responses

| References | Population | Perturbation characteristics | | | |
|---|---|---|---|---|---|
| | | Directions | Conditions | Preload | Energy |
| Load-dropping studies | | | | | |
| Foust [43] | Healthy adults. Females, $N=93$; males, $N=77$ (45%). Young adult (18–24 years), middle-age (35–44 years), elderly (62–74 years) | Flexion, extension | Unanticipated | 57 g | 0.45–0.89 J (455 g from 10 to 20 cm depending on the participant) |
| Reid et al. [30] | Young adults. Females, $N=1$; males, $N=7$ (87%). 18–23 years | 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 to 100 cm) |
| Corna et al. [42] | 'Normal' participants. Females, $N=3$; males, $N=7$ (70%). 22–49 years ($N=7$ '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, 2 kg from < 3 mm) |
| Mansell et al. [32] | NCAA collegiate soccer players. Females, $N=19$, 19.16 ± 0.90 years; males, $N=17$ (47%), 19.21 ± .92 years | Flexion, extension (randomized) | Anticipated, unanticipated (in that sequence) | Weight of cable and landing surface, not given | 1.47 J (1 kg from 15 cm) |
| Tierney et al. [50] | Physically active population. Females, $N=20$, 24.2 ± 4.1 years; males, $N=20$ (50%), 26.3 ± 4.3 years | 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. [44] | Young adults. Males, 10 (100%), 24.0 ± 5.2 years | 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 (3 kg from 40 cm landing on a spring) |
| Simoneau et al. [31] | Healthy adults. Females, $N=3$; males, $N=4$ (57%). 23.5 years | 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. [29] | Contact-sport athletes (soccer, ice hockey, US football, martial arts, wrestlers, lacrosse). Females, $N=22$, 15.0 ± 4.4 years; males, $N=24$ (52%), 16.3 ± 5.0 years. 8–30 years | 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 counterbalanced by an opposite spring | 0.15–0.78 J (1 kg from 1.5 to 8 cm, depending on BW) |
| Schmidt et al. [33] | High-school and collegiate football players. Males, $N=49$ (100%), 18.55 ± 1.15 years | Flexion, extension (randomized) | Anticipated, unanticipated (in that sequence) | 1% BW | 3.21–3.97 J (2.5% BW mass from 15 cm) |

**Table 1** (continued)

| References | Population | Perturbation characteristics | | | |
|---|---|---|---|---|---|
| | | Directions | Conditions | Preload | Energy |
| Debison-Larabie [54] | Ice hockey players, Varsity and competitive leagues. Females, $N=8$, 20.60 ± 1.30 years; males, $N=8$ (50%), 22.13 ± 1.55 years | 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. [51] | Recreationally active young adults. Females $N=10$; males $N=9$ (47%). 22.5 ± 1.7 years (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 |
| Quick-release studies | | | | | |
| Ito et al. [45] | 'Normal' adults. Females $N=2$; males $N=8$ (80%). 21–49 years<br>LDP. Males $N=4$ (100%). 29–34 years | Extension (supine) | Unanticipated | No preload | Not applicable |
| Corna et al. [42] | 'Normal' participants. Females $N=3$; males $N=7$ (70%) [range 22–49 years]<br>LDP. Females $N=2$; males $N=4$ (67%). 35–73 years<br>($N=10$ 'normal' participants, with 4 LDP among these) | Flexion, rotation | Unanticipated, during movement | Females: 2 kg<br>Males: 3 kg | Not applicable |
| Ito et al. [46] | 'Normal' adults. Females $N=3$; males $N=7$ (70%). 23—49 years<br>LDP. Females $N=2$; males $N=4$ (67%). 29–68 years | Extension (supine) | Unanticipated, with passive motion or active resisting (in that sequence) | No preload | Not applicable |
| Portero et al. [52] | Healthy adults. Females $N=2$; males $N=8$ (80%). 30.6 years | Flexion, extension (randomized) | Unanticipated release | 6 submaximal isometric force levels (from 20 to 70% MVC) | Not applicable |
| Portero et al. [49] | Healthy adults. Females $N=2$; males $N=11$ (85%). 27.1 ± 3.2 years | Flexion, extension (randomized) | Unanticipated | 8 submaximal isometric force levels (from 10 to 80% MVC) | Not applicable |
| Portero et al. [48] | Healthy adults. Females $N=2$; males $N=6$ (75%). 31.7 ± 2.6 years | Flexion | Unanticipated release | Submaximal isometric force levels (from 20 to 70% MVC) | Not applicable |

**Table 1** (continued)

| References | Population | Perturbation characteristics: Directions | Conditions | Preload | Energy |
|---|---|---|---|---|---|
| Direct-impact studies | | | | | |
| Kuramochi et al. [47] | Healthy adults. Males, $N=9$ (100%), 22.6 ± 4.1 years (19–30) | Extension | Anticipated, unanticipated (randomized) | No preload, no weight | 1.66 J (pendulum of 4 kg on a 45 cm, 25° string) |
| Lucas [55] | Females $N=1$, 29 years; males $N=1$ (50%), 23 years | Flexion, extension (randomized) | Unanticipated | No preload, no weight | Not reported (maximum force transmitted to the head: 12.7 ± 1.2 N) |
| Vasavada et al. [53] | Females $N=5$. Males $N=4$ (44%), age not given | Extension, 45° rotated extension | Unanticipated | No preload, no weight | Not reported |

Data are expressed as 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

or presented details on the apparatus [55] used in subsequent research [53].

All studies were cross-sectional experimental studies except for one cohort study [33]. 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 (see electronic supplementary Table S1).

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 [30, 42, 43, 45, 46]. One Master's thesis [55] and one conference abstract [53] 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.

## 3.2 Methodologies

Methodological approaches utilising a mechanical perturbation to the head to assess head/neck dynamic responses (see Table 1) included load dropping (11 publications) [29–33, 42–44, 50, 51, 54], quick release (6 publications) [42, 45, 46, 48, 49, 52], direct impact to the head via a pendulum [47], and a motorized impactor [53, 55]. One study [42] reported outcome measurements for both load-dropping and quick-release methods.

### 3.2.1 Load Dropping

Eleven studies [29–33, 42–44, 50, 51, 54] involved the sudden dropping of a load to induce head perturbation (see Fig. 2). This methodology was utilised with athlete populations [29, 32, 33, 50, 54] to investigate head/neck dynamic responses in traffic accidents (i.e. whiplash injury research with healthy adults) [43, 44] or to examine human neck reflex mechanisms [30, 31, 42, 51].

The load-dropping perturbation occurred by the impact of a free-falling weight on a landing surface that was connected to the head via a non-extensible cable and a head harness [29, 30, 32, 33, 50, 51, 54], a headband or strap [31, 43, 44], or a plate held between the teeth [42] (see Fig. 2). The cable

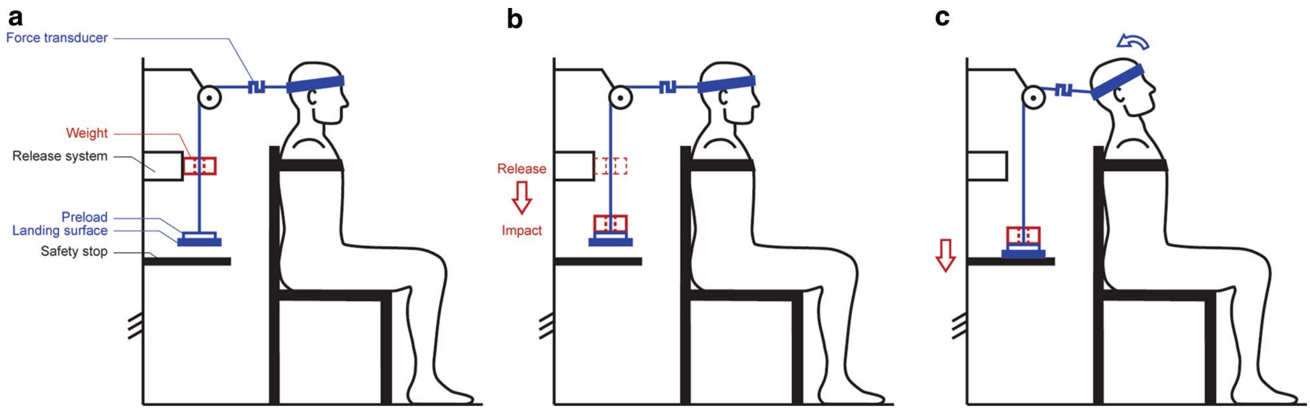


**Fig. 2** Load-dropping method on neck extension. **a** The participant sustains a generally static light load, with an optional preload mass. **b** The weight is released and falls onto the landing surface. **c** The impact created pulls on the participant's head

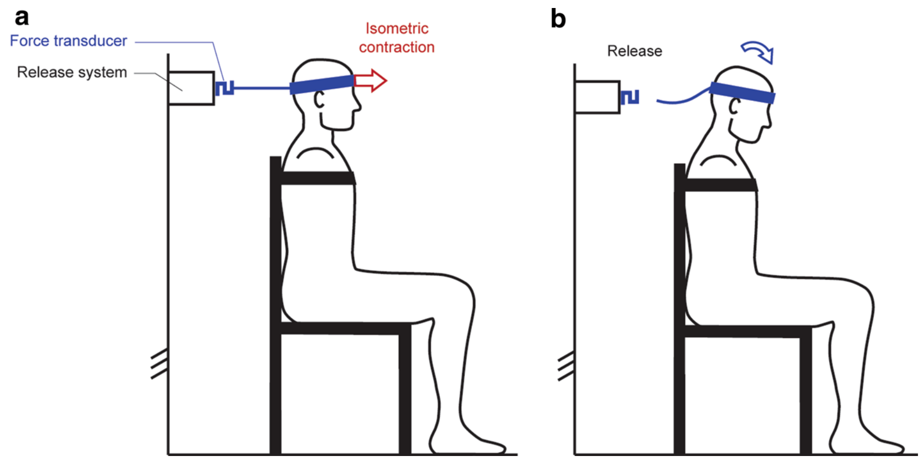


**Fig. 3** Illustration of the quick-release method in flexion as described by Portero et al. [48, 49, 52]. **a** Isometric contraction followed by **b** the sudden and unexpected release of the cable leads to forward head motion

ran through a height-adjustable pulley to ensure the force was applied perpendicular to the head/neck segment. The weight was either manually released [33] or via an electromagnet [31, 42, 43, 54]. Only two studies utilised a safety stop to limit the displacement of the head set at 2.5 [29] or 10 cm [44] from its initial position. All load-dropping studies investigated participant's responses in a seated position. Participants were static before the perturbation application in the majority of studies, while in one study [42] participants were actively moving a weight.

#### 3.2.2 Quick Release

Six publications [42, 45, 46, 48, 49, 52] reported the use of quick-release methods for the assessment of head/neck segment dynamic properties. Testing conditions involved relaxed neck muscles [45, 46], isometric submaximal contraction [48, 49, 52], or a loaded dynamic movement [42]. In each case, the head was attached via a cable that was suddenly, and unexpectedly, released (see Figs. 3, 4).

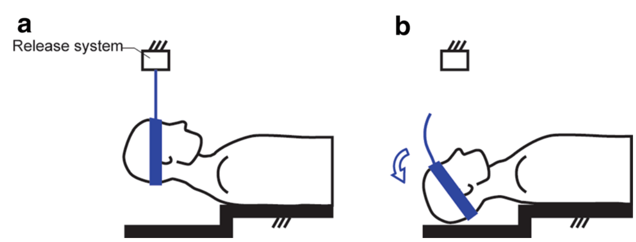


**Fig. 4** Illustration of the quick-release method as described by Ito et al. [45, 46]. **a** Participants lay supine with their head resting in a sling in a slightly flexed position. **b** The sling is released and the head free falls to an extended position onto a cushioned landing surface

Participants were healthy [48, 49, 52] or presented labyrinthine deficiency [42, 45, 46].

#### 3.2.3 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 [47], or with a linear motor that impulsed a hit at the vertex of a helmet worn by participants [53, 55].

## 3.3 Factors to Consider for Assessment of Head/Neck Dynamic Responses

### 3.3.1 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 [51, 54], of the studies included. Two studies [51, 54] utilised a second investigator to check body position throughout the assessment. However, it has been reported [56] 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.

### 3.3.2 Instructions

As is often the case in experiments involving human participation, variability can arise from the instructions given to participants [57]. 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' [32], 'as soon as', [50] or 'once they feel the tug' [33, 51], 'maintain the head still' [31], or 'right their head' and 'resume tracking as quickly as possible' [42, 46]. Analysis of the effects of different instructions on reaction times and computed neck stiffness [30] 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 with 'resist as much as desired' [30].

Other studies [42, 45–47, 53, 55] assessed passive motion, where participants moved freely [53] with the perturbation without resisting. Passive motion, compared with active resisting, showed greater head displacement and peak velocity, as well as reduced muscular activity in healthy participants [42, 46]. In the active quick-release studies by Portero et al. [48, 49, 52], the authors reportedly chose to study head movement in the first 15–30 ms, preceding reflex and voluntary muscle reactions, and thus suppressing the need for instructions. The use of instructions was not reported in the other studies [29, 43, 44, 54].

Researchers should be aware of the variability induced by instructions, and choose to study active resisting or passive motion depending on 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 [42, 46]. 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.

### 3.3.3 Preloading

Included studies reported that the amount of preloading influences head/neck response [31, 48, 49, 51, 52]. Therefore, depending on the load, 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 [31], 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 participants.

### 3.3.4 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 [58]. Bracing for the impact allows an anticipatory co-contraction of the neck muscles to help reduce the consequences of the perturbation [31]. Therefore, depending on the sport or situation being investigated, studying head/neck responses to both anticipated and unanticipated perturbations is recommended. All but one study [44] reported the participant's level of anticipation. All the quick-release [42, 45, 46, 48, 49, 52] and four load-dropping tests [31, 42, 43, 53] were performed with participants systematically unaware of the onset of the perturbation. Other studies [29, 30, 32, 47, 50, 51, 54] explored the effects of anticipation on head and neck responses (results are summarized in Table 2 and in Sect. 3.5.4), except for one study [33] 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 participants safely accommodate the impact [32, 33, 47, 50]. It is hypothesized that most experiments performing anticipated trials systematically before unanticipated trials chose to ensure participants' safety and comfort by doing so. However, this familiarization might bias the unanticipated behaviour as the perturbation level is not completely surprising [50].

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 [30, 42]. Therefore, noise and visible movement produced by the devices should be limited. Several protocols actively blocked visual and auditory cues by closing the eyes [45–47, 54], wearing blackened goggles [32, 33, 50, 53], masking the perturbation device [43], and/or noise-cancelling devices [32, 33, 47, 50, 53].

#### 3.3.5 Analysis of Multiple Trials

Most studies included in this review reported that participants performed at least three trials for each condition and averaged the results of all trials. Some studies discarded the data from the first trial [33, 51] or the first three trials [55] of each series "to eliminate a possible exaggerated neuromuscular startle response during the first exposure". Indeed, in the first unanticipated trial, especially when there have been no previous anticipated trials, a 'startle' phenomenon can occur [59]. Siegmund et al. [59] 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 muscle activity between the first and subsequent exposures [60, 61]. The reduction of the startle phenomenon with repeated exposures is called habituation [60]. Although it has never been studied in experiments with direct head perturbation, four of the reviewed studies [30, 32, 45, 47] mentioned the habituation phenomenon. Reid et al. [30] measured a drop of 50% in the stiffness between the first and second trial and attributed this to participants realizing that the tug was not harmful and that they could relax. Ito et al. [45] noticed greater muscular activity for the first trial, as well as a more pronounced eye blink, while Mansell et al. [32] measured 40% differences in head accelerations and muscle onset latency between two test sessions separated by approximately 10 weeks. In contrast, Kuramochi et al. [47] 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. However, excluding the first trial could rule out useful information [59, 61], 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 trial 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.

#### 3.3.6 Body Position, Direction and Location of Force Application

Most, but not all [45, 46, 53], 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 [62]. Studies have reported that body [63] and head position [64, 65] influence neck force production. For validity purposes, future research should capitalize on head impact 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 [30, 31, 33, 43, 44, 51, 54] applied the perturbation to the forehead for flexion, or occiput for extension, some [32, 50, 53] chose to apply it to the vertex of the head or had participants holding a plate between their teeth [42]. However, Fukushima et al. [44] 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 [30, 44]. 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 [44]. This phenomenon, termed 'cervical retraction' [44, 66], is even more pronounced when the perturbation is applied at the maxilla level when compared with a forehead application; therefore, it can be expected that the results would be different if applied at the vertex.

#### 3.3.7 Mechanical Loading

The energy of the impacts was not reported in studies utilising the load-dropping [29–33, 42–44, 50, 54] and direct-impact [47, 53] methods. One study [51] 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 $<0.1$ J to potentially 24.5 J (see Table 1). The majority (69%) of the studies [30–32, 42, 44, 47, 50, 53, 54] utilised the same energy for every participant, while other studies varied the weight [29, 33] or fall height [43, 51] based on participants' body weight. Only four studies [32, 47, 50, 54] 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 [32, 50], and one study [47] 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 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 [29, 44]. This includes shock absorption properties of the impacting components [43, 44], tension of the cable, stretchiness and fitting of the headpiece [29]. Dampening of the perturbation modifies the force that is transmitted to the head, and most likely influences head kinematics [44]. 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) [62], 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 [31–33, 50, 67]. However, this was only partially described in three studies [32, 44, 50], with the impact force reported to average 50 [32, 50] to 200 N [44]. 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 interstudy 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, time to peak force, and variability of these values.

## 3.4 Magnitude, Reliability and Variability of Head/Neck Dynamic Response Variables

### 3.4.1 Magnitude

Eighteen different metrics have been utilised to report head/neck dynamic responses to perturbation, with the most frequent being neck muscle latency (nine studies, 18.6–88.0 ms) [32, 33, 42, 43, 45, 46, 50, 51, 54], neck stiffness (seven studies, 147.2–721.9 N/rad, 14–1145.3 Nm/rad) [31–33, 48–50, 52], and linear head acceleration (four studies, 0.2–3.8*g*) [42, 45–47].

Most, but not all [47, 48, 51, 52], studies recorded head kinematics using two-dimensional [32, 33, 50] or three-dimensional motion capture [29, 53, 54], accelerometers [30, 42, 43, 45, 46, 49], angular velocity sensors [31], or cineradiography [44]. The reported head kinematic variables varied greatly across studies and included linear and/or angular peak accelerations, decelerations, velocities or displacements and time to peak acceleration.

Sampling frequency of the included studies varied between 60 and 1500 Hz. However, Ito et al. [46] 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 [68]. Furthermore, for real-time head impact measurements, it has been recommended [69] 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 variables was infrequently justified, but some kinematic metrics have been suggested to be associated with concussion [70–72], occurrence of whiplash injuries [54], or brain tissue deformation [29, 73]. Linear and angular motions have different effects on brain injury mechanisms [15] and the movement caused by a direct perturbation to the head is a combination of translation and rotation [44]. 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 [29].

Several composite injury metrics have also been proposed, associating several variables [74, 75], 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 studies of head impacts [67, 76], there is no individual metric that is unequivocally accepted as being associated with brain injuries [77–79].

Experiments often included surface electromyography (EMG) for the measurement of muscle activity [30, 32, 33, 42, 43, 45–47, 50, 51, 53–55]. A few studies did not record muscular activity by EMG [29, 31, 48, 49, 52], and one study [44] used EMG to control the participant's relaxed state before perturbation. Authors rarely reported peak [32, 50, 51] or mean [32, 47, 50, 51, 54] muscle activity, while all papers presented muscle onset latencies. Latencies were calculated utilising various methods, such as a change of magnitude by visual analysis [30, 43] or the precise calculation of activity threshold [33, 51].

Each of the seven studies that investigated musculotendinous stiffness utilised either variations of Eq. (1) [30, 32, 33, 48–50, 52], or a more complex mathematical model to compute stiffness and viscosity [31]. Equation (1) identifies that $F$ is the force applied on the body and $\delta$ is the displacement produced by the force.

$$k = \frac{F}{\delta} \tag{1}$$

Despite reasonably similar protocols [31–33, 50], the comparison of stiffness results across studies is limited. Stiffness was reported using either force [32, 50] or torque [31, 33] measurements, and was normalized for some [33], but not all [31, 32, 50], studies by participants' body weight.

A few studies normalized head kinematics [29] or neck stiffness [33] by participants' body weight [29, 33] and height [33], presuming there is a relationship between stiffness and body characteristics. This relationship is explained by the mathematical model utilised by Simoneau et al. [31] that describes stiffness as a generator of torque acting against gravity and perturbation forces [31]. 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 [31, 32, 44, 50]. Data normalization reduces intersubject variability, and it is therefore recommended that future studies normalise data, especially when comparing populations.

#### 3.4.2 Reliability

Only two studies [49, 50] reported the reliability of the perturbation methodology providing test–retest results. Tierney [50] reported intraclass correlation coefficients (ICCs) of 0.98 (95% confidence interval [CI] 0.72–0.92) for head kinematics, 0.92 for neck muscle peak activity, 0.72 for muscle onset latency, and 0.96 for force measurement with the load-dropping method. Portero et al. [49] reported an ICC of 0.81–0.96 and a standard error of measurement (SEM) of 0.9–2.2 Nm/deg for neck stiffness with the quick-release method.

#### 3.4.3 Validity

Over one football season, one study [33] compared the associations between head/neck response to perturbation and real head impacts measured using the Head Impact Telemetry System (HITS). It was reported that increased stiffness of the flexor muscles was associated with reduced odds of sustaining moderate and severe head impacts (odds ratio [OR] 0.68, 95% CI 0.48–0.96). Reactivity in the flexor muscles, i.e. 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) [33]. Being able to limit head angular displacement during perturbation testing has limited the effects on the odds of sustaining higher magnitude impacts during sports participation (no ORs reached statistical significance) [33].

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 [16], either by another participant [80–82], a moving object [83], or the environment [58]. Pendulum [47] or motors [53, 55] 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 recreate 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 [45]. 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, when compared with the direct-impact methods, the load-dropping 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 [42, 45, 46, 48, 49, 52]. Specifically, the quick-release methods utilised by Portero et al. [48, 49, 52] focused on characterizing passive musculotendinous stiffness during the 15–30 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 [42, 45, 46] 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 [42]. These studies identified that vestibular and stretch reflexes exhibited approximately 25 ms and approximately 65 ms latencies, respectively [42, 45, 46]. Therefore, the quick-release methodologies [48, 49, 52] 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 [33] based on the sport that was investigated. However, real-life head impacts rarely follow the anatomical planes [84], 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. [33] 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 [72, 85, 86]. Additional research is warranted to determine which neck muscles are involved in counteracting 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 [15] that rotational accelerations play a major role in concussion injury, future experiments focusing on rotational movement would provide 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 5777 rad/s$^2$ [76]. Examination of the studies included in this review showed that the linear and angular accelerations reported did not exceed 4 *g* [29, 42, 45–47] and 42 rad/s$^2$ [32, 50, 54], or 4.2% and 0.7% of the estimated mean concussive accelerations, respectively. In addition, 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 [79]. 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 [76], there is currently no consensus that this is the most appropriate variable to describe head/neck responses to real-life head impact with regard to concussion risk [71, 87]. 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.

### 3.5 Covariates 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 (44–100% males), activity level (competitive athletes to non-sporting participants), age (8–74 years), and conditions of perturbation application such as anticipation and preloading (see Table 1). The effects of these covariates are reported in Table 2 for the 13 studies that reported information on covariates. Studies not shown in Table 2 [30, 42, 44, 45, 53, 55] did not report any analysis of the covariate effects.

#### 3.5.1 Sex

Some studies [43, 50, 54] reported males had slower reflex times when compared with females. It was also reported that female muscles were able to start contracting more promptly, although this is not corroborated for all directions of perturbation [54]. Females also seem to be able to decelerate their head motion faster and stronger than males [43]. However, it has been reported [50, 54] 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 [29]. 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 with men [50]. Reduced neck strength and stiffness were also reported for females, and may be related to their smaller amount of muscular tissue [50, 54]. Mansell et al. [32] did not observe any sex-related differences, and attributed this to the level of physical activity of participants. 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 [32] suggested this would have reduced the differences between the sexes.

To date, there is no consensus on sex as a risk factor for concussion [19], but there is evidence that sex differences do exist in the outcomes of concussion [88]. 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 they should be considered as a modifying factor for head/neck dynamic response to perturbation.

**Table 2** Effects of covariates on head/neck dynamic response to perturbation in 13 studies that reported covariates such as sex, age, anthropometrics, anticipation, preloading, peak or rate of force/torque

| Covariate; study, method | Effects of the covariate |
|---|---|
| Sex | |
| Debison-Larabie [54], 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. [29], load dropping | No effect on peak lin and ang vel |
| Foust [43], load dropping | Females have ↓ latencies, ↑ head deceleration, ↓ time to peak deceleration |
| Mansell et al. [32], load dropping | No sex differences in kinematics, EMG or stiffness |
| Tierney et al. [50], load dropping | Females have ↑ ang acc, ↑ ang disp, ↑ muscle activity (peak and area), ↓ latencies (SCM, trap in ext), ↓ stiffness |
| Age | |
| Eckner et al. [29], load dropping | Effect of age on peak lin and ang vel ($p<0.001$) |
| Foust [43], load dropping | Elderly age group has ↑ muscle latencies, ↑ time to peak deceleration |
| Ito et al. [46], quick release | No correlation between age and muscle latencies |
| Anthropometrics | |
| Debison-Larabie [54], load dropping | Weak to no relationship between HC/NC, neck volume or TNV and head ang acc |
| Eckner et al. [29], load dropping | ↑ CSA (SCM) = ↓ lin and ang vel in extension, ↑ NC = ↓ lin and ang vel for all directions |
| Foust [43], load dropping | No effect of stature |
| Schmidt et al. [33], load dropping | All players, larger SCM and SSC = ↑ odds[a] |
| Alsalaheen et al. [51], load dropping | No association between NC or SCM CSA and neuromuscular response |
| Isometric peak force/torque | |
| Eckner et al. [29], 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. [33], load dropping | All players: equal odds[a] between high and low performers (lin acc and $HIT_{SP}$). For linemen, stronger Lflex and comp: ↑ odds[a] |
| Alsalaheen et al. [51], load dropping | No association between peak force (flex) and neuromuscular response |
| Rate of force/torque development | |
| Eckner et al. [29], load dropping | ↓ Lin vel for ext, flex ($p<0.05$); ↓ ang vel for ext, flex, rot ($p<0.05$) |
| Schmidt et al. [33], load dropping | All players, higher ext RTD: ↑ odds of sustaining severe impacts. For skill players, higher flex, ext, Lflex, comp RTD: ↑ odds[a] |
| Alsalaheen et al. [51], load dropping | No association between RFD (flex) and neuromuscular response |
| Anticipated compared with unanticipated | |
| Debison-Larabie [54], load dropping | ↓ Muscular activity in Lflex (1.4%) |
| Eckner et al. [29], load dropping | ↓ Head lin (12.3%) and ang (9.7%) vel across all directions ($p<0.001$) |
| Mansell et al. [32], load dropping | ↓ Ang displacement (ext: 23%, flex: 25%), ↓ SCM peak activity (18%) |
| Tierney et al. [50], load dropping | Males only: ↓ ang acc (25%) |
| Alsalaheen et al. [51], 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. [46], quick release | ↓ Peak head lin vel |
| Kuramochi et al. [47], direct impact | ↓ Muscular activity (SCM, ext) |
| Preloading | |
| Portero et al. [49], quick release | ↑ Head ang disp ($0.94<R^2<0.99$) |
| Portero et al. [52], quick release | ↑ Stiffness ($0.45<R^2<0.68$) |
| Portero et al. [48], quick release | ↑ Stiffness ($R^2=0.74$) |
| Simoneau et al. [31], 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. [51], 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 |

*HC* head 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, ↑ indicates increased, ↓ indicates decreased

[a]Odds of sustaining moderate or severe head impacts compared with mild head impacts [33]

#### 3.5.2 Age

There appears to be an age effect on neck strength and head/neck response to perturbation. Younger (high school or younger) [29] and elderly [43] 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 18–50 years of age [43, 46]. There is a potential interaction between sex and age, with the strength capacities of adult males and females peaking in the middle-age and young adult categories, respectively [43].

#### 3.5.3 Anthropometrics, Neck Strength and Force Development

Stature did not influence head/neck dynamic response [43]. Head and neck dimensions, including muscle volume, might have a slight effect on head kinematics [29, 54] but not on neuromuscular response [51]. However, it seems that larger cervical muscles decrease perturbation kinematics [29], but increase the odds of sustaining high-magnitude head impacts during football play [33].

Six studies [29, 32, 33, 43, 50, 51] assessed isometric neck strength or rate of force development (RFD). These parameters were tested for their effects on the dynamic response to perturbation. There were significant correlations ($r = 0.417$–$0.657$) between neck strength and linear and angular head velocities in most directions of perturbation [29]. Other studies [32, 43, 50] 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. [50], 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 other studies [32, 43]. Corroborating the absence of relationship, in an intervention study of 36 soccer players [32], a significant increase (15%, $p < 0.001$) 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 [51]. Furthermore, in the only study [33] 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 with players with weaker, thinner necks (ORs ranging from 0.88 to 1.65) [33].

While there does not seem to be an association between RFD and neck neuromuscular response [51], Eckner et al. [29] identified that a higher RFD reduced head kinematics in several directions. This is further illustrated by two studies utilising the same perturbation protocol, which showed that trained soccer players [32] exhibited less head angular acceleration than physically active participants [50]. 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 [33] or ice hockey players [54]. 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 [33].

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 [32, 33, 37] 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, as well as fatigue on neck muscle capacities, and how these impact on the response to perturbations and the risk of concussions.

#### 3.5.4 Anticipation Conditions in the Studies

The effects of anticipation on head motion are contradictory, with studies reporting similar [32, 47, 54] or decreased [29, 32, 50] head acceleration, velocity, and displacement when compared with unanticipated perturbations. Neck stiffness showed no significant ($p > 0.05$) difference between anticipation conditions [32, 50]. Neck muscle activity was reduced when participants braced for the impact in most [32, 47, 54], but not all [50, 51], studies. Muscle onset latency was reduced [51], as well as being more efficient [30] (i.e. meaning the participant postponed muscle activity until immediately prior to the onset of the perturbation) when participants knew of the perturbation onset [30]. 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.

#### 3.5.5 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 the landing surface [30, 32, 42–44, 50, 51, 54]. A spring was utilised in one study [29] to counterbalance 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 [31, 33, 51] and study its effects on head kinematics, neck viscoelastic properties [31] and neuromuscular response [51]. It has been reported [31] that pre-perturbation loading is positively associated with increased muscular activity [31, 51], neck stiffness and viscosity [31], and decreased head kinematics [31] and muscle onset latency [51]. These results are further supported by quick-release experiments [48, 52] where neck stiffness was dependent on the torque applied before release.

## 4 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 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 perturbation, making the calculation of neck stiffness challenging. For this reason, the load-dropping method is recommended.

The magnitude of perturbation should be kept under 4 J for the safety of participants 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 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 covariates.

When undertaking research, it is recommended that age should be a consideration, especially if the population is outside of the 18–50 years 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 regard 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, to our knowledge, this has only been shown in one study [33]. 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 [29, 33, 51]. Further research is warranted to identify head/neck dynamic response variables related to injury risk.

In summary, each methodology can provide useful information on the head/neck dynamic response. However, the validity and relevance of these methodologies when compared with in vivo impact measurement still needs to be addressed. Reports on head/neck response should include neck muscle latency (ms), neck stiffness (N/rad or Nm/rad) 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.

**Conflict of interest** 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.

## References

1. Fuller CW, Brooks JH, Cancea RJ, Hall J, Kemp SP. Contact events in rugby union and their propensity to cause injury. Br J Sports Med. 2007;41(12):862–7.
2. Willigenburg NW, Borchers JR, Quincy R, Kaeding CC, Hewett TE. Comparison of injuries in American collegiate football and club rugby: a prospective cohort study. Am J Sports Med. 2016;44(3):753–60.

3. Quarrie KL, Hopkins WG. Tackle injuries in professional Rugby Union. Am J Sports Med. 2008;36(9):1705–16. https://doi.org/10.1177/0363546508316768.
4. Dick R, Ferrara MS, Agel J, Courson R, Marshall SW, Hanley MJ, et al. Descriptive epidemiology of collegiate men's football injuries: National Collegiate Athletic Association Injury Surveillance System, 1988–1989 through 2003–2004. J Athl Train. 2007;42(2):221–33.
5. Swain MS, Lystad RP, Pollard H, Bonello R. Incidence and severity of neck injury in Rugby Union: a systematic review. J Sci Med Sport. 2011;14(5):383–9.
6. Clay MB, Glover KL, Lowe DT. Epidemiology of concussion in sport: a literature review. J Chiropr Med. 2013;12(4):230–51.
7. Brauge D, Delpierre C, Adam P, Sol JC, Bernard P, Roux F-E. Clinical and radiological cervical spine evaluation in retired professional rugby players. J Neurosurg Spine. 2015;23(5):551–7.
8. Triantafillou KM, Lauerman W, Kalantar SB. Degenerative disease of the cervical spine and its relationship to athletes. Clin Sports Med. 2012;31(3):509–20. https://doi.org/10.1016/j.csm.2012.03.009.
9. Hume PA, Theadom A, Lewis GN, Quarrie KL, Brown SR, Hill R, et al. A comparison of cognitive function in former rugby union players compared with former non-contact-sport players and the impact of concussion history. Sports Med. 2017;47(6):1209–20. https://doi.org/10.1007/s40279-016-0608-8.
10. Manley G, Gardner AJ, Schneider KJ, Guskiewicz KM, Bailes J, Cantu RC, et al. A systematic review of potential long-term effects of sport-related concussion. Br J Sports Med. 2017;51(12):969–77. https://doi.org/10.1136/bjsports-2017-097791.
11. Rihn JA, Anderson DT, Lamb K, Deluca PF, Bata A, Marchetto PA, et al. Cervical spine injuries in American football. Sports Med. 2009;39(9):697–708. https://doi.org/10.2165/11315190-000000000-00000.
12. Michael DB, Guyot DR, Darmody WR. Coincidence of head and cervical spine injury. J Neurotrauma. 1989;6(3):177–89. https://doi.org/10.1089/neu.1989.6.177.
13. Cheever K, Kawata K, Tierney R, Galgon A. Cervical injury assessments for concussion evaluation: a review. J Athl Train. 2016;51(12):1037–44. https://doi.org/10.4085/1062-6050-51.12.15.
14. Hynes LM, Dickey JP. Is there a relationship between whiplash-associated disorders and concussion in hockey? A preliminary study. Brain Inj. 2006;20(2):179–88. https://doi.org/10.1080/02699050500443707.
15. King AI, Yang KH, Zhang L, Hardy W, Viano DC (eds). Is head injury caused by linear or angular acceleration. In: International research council on biomechanics of Injury (IRCOBI) conference; 2003.
16. Delaney JS, Puni V, Rouah F. Mechanisms of injury for concussions in university football, ice hockey, and soccer: a pilot study. Clin J Sport Med. 2006;16(2):162–5.
17. Winkelstein BA, Myers BS. The biomechanics of cervical spine injury and implications for injury prevention. Med Sci Sports Exerc. 1997;29(7 Suppl):S246–55.
18. Dennison CR, Macri EM, Cripton PA. Mechanisms of cervical spine injury in rugby union: is it premature to abandon hyperflexion as the main mechanism underpinning injury? Br J Sports Med. 2012;46(8):545–9.
19. McCrory P, Meeuwisse W, Dvorak J, Aubry M, Bailes J, Broglio S, et al. Consensus statement on concussion in sport—the 5th International Conference on Concussion in Sport held in Berlin, October 2016. Br J Sports Med. 2017;51(11):838–47. https://doi.org/10.1136/bjsports-2017-097699.
20. Gennarelli T, Segawa H, Wald U, Czernicki Z, Marsh K, Thompson C. Physiological response to angular acceleration of the head. Head injury: basic and clinical aspects. 1982;1982:129–40.
21. Mainwaring L, Ferdinand Pennock KM, Mylabathula S, Alavie BZ. Subconcussive head impacts in sport: a systematic review of the evidence. Int J Psychophysiol. 2018. https://doi.org/10.1016/j.ijpsycho.2018.01.007.
22. Moon DW, Beedle CW, Kovacic CR. Peak head acceleration of athletes during competition-football. Med Sci Sports. 1971;3(1):44–50.
23. Duma SM, Manoogian SJ, Bussone WR, Brolinson PG, Goforth MW, Donnenwerth JJ, et al. Analysis of real-time head accelerations in collegiate football players. Clin J Sport Med. 2005;15(1):3–8.
24. King D, Hume PA, Brughelli M, Gissane C. Instrumented mouthguard acceleration analyses for head impacts in amateur rugby union players over a season of matches. Am J Sports Med. 2015;43(3):614–24. https://doi.org/10.1177/0363546514560876.
25. Panjabi MM, Cholewicki J, Nibu K, Grauer J, Babat LB, Dvorak J. Critical load of the human cervical spine: an in vitro experimental study. Clin Biomech. 1998;13(1):11–7. https://doi.org/10.1016/S0268-0033(97)00057-0.
26. Brault JR, Siegmund GP, Wheeler JB. Cervical muscle response during whiplash: evidence of a lengthening muscle contraction. Clin Biomech. 2000;15(6):426–35. https://doi.org/10.1016/S0268-0033(99)00097-2.
27. Kumar S, Narayan Y, Amell T. Role of awareness in head-neck acceleration in low velocity rear-end impacts. Accid Anal Prev. 2000;32(2):233–41.
28. Collins CL, Fletcher EN, Fields SK, Kluchurosky L, Rohrkemper MK, Comstock RD, et al. Neck strength: a protective factor reducing risk for concussion in high school sports. J Prim Prev. 2014;35(5):309–19. https://doi.org/10.1007/s10935-014-0355-2.
29. Eckner JT, Oh YK, Joshi MS, Richardson JK, Ashton-Miller JA. Effect of neck muscle strength and anticipatory cervical muscle activation on the kinematic response of the head to impulsive loads. Am J Sports Med. 2014;42(3):566–76. https://doi.org/10.1177/0363546513517869.
30. Reid SE, Raviv G, Reid SE Jr. Neck muscle resistance to head impact. Aviat Space Environ Med. 1981;52(2):78–84.
31. Simoneau M, Denninger M, Hain TC. Role of loading on head stability and effective neck stiffness and viscosity. J Biomech. 2008;41(10):2097–103. https://doi.org/10.1016/j.jbiomech.2008.05.002.
32. Mansell J, Tierney RT, Sitler MR, Swanik KA, Stearne D. Resistance training and head-neck segment dynamic stabilization in male and female collegiate soccer players. J Athl Train. 2005;40(4):310–9.
33. Schmidt JD, Guskiewicz KM, Blackburn JT, Mihalik JP, Siegmund GP, Marshall SW. The influence of cervical muscle characteristics on head impact biomechanics in football. Am J Sports Med. 2014;42(9):2056–66. https://doi.org/10.1177/0363546514536685.
34. Kumar S, Ferrari R, Narayan Y. Kinematic and electromyographic response to whiplash loading in low-velocity whiplash impacts—a review. Clin Biomech. 2005;20(4):343–56.
35. Yoganandan N, Pintar FA. Biomechanics of human head-neck in rear impacts. Int J Vehicle Des. 2003;32(1–2):53–67. https://doi.org/10.1504/Ijvd.2003.003236.
36. Mihalik JP, Guskiewicz KM, Marshall SW, Greenwald RM, Blackburn JT, Cantu RC. Does cervical muscle strength in youth ice hockey players affect head impact biomechanics? Clin J Sport Med. 2011;21(5):416–21.
37. Gilchrist I, Storr M, Chapman E, Pelland L. Neck muscle strength training in the risk management of concussion in contact sports: critical appraisal of application to practice. J Athl Enhanc. 2015;4(2):19.

38. Moher D, Liberati A, Tetzlaff J, Altman DG. Preferred reporting items for systematic reviews and meta-analyses: the PRISMA statement. Ann Intern Med. 2009;151(4):264–9.
39. von Elm E, Altman DG, Egger M, Pocock SJ, Gøtzsche PC, Vandenbroucke JP. The Strengthening the Reporting of Observational Studies in Epidemiology (STROBE) statement: guidelines for reporting observational studies. The Lancet. 2007;370(9596):1453–7. https://doi.org/10.1016/S0140-6736(07)61602-X.
40. Vandenbroucke JP, Von Elm E, Altman DG, Gøtzsche PC, Mulrow CD, Pocock SJ, et al. Strengthening the reporting of observational studies in epidemiology (STROBE): explanation and elaboration. PLoS medicine. 2007;4(10):e297.
41. Popay J, Roberts H, Sowden A, Petticrew M, Arai L, Rodgers M, et al. Guidance on the conduct of narrative synthesis in systematic reviews. A product from the ESRC methods programme version. 2006;1:b92.
42. Corna S, Ito Y, von Brevern M, Bronstein AM, Gresty MA. Reflex (unloading) and (defensive capitulation) responses in human neck muscle. J Physiol. 1996;496(Pt 2):589–96.
43. Foust DR. Cervical range of motion and dynamic response and strength of cervical muscles. In: Proceedings of the 17th stapp car crash conference; 17–19 Nov 1973: Coronado, CA.
44. Fukushima M, Kaneoka K, Ono K, Sakane M, Ujihashi S, Ochiai N. Neck injury mechanisms during direct face impact. Spine. 2006;31(8):903–8. https://doi.org/10.1097/01.brs.0000209257.47140.fc.
45. Ito Y, Corna S, von Brevern M, Bronstein A, Rothwell J, Gresty M. Neck muscle responses to abrupt free fall of the head: comparison of normal with labyrinthine-defective human subjects. J Physiol. 1995;489(Pt 3):911–6.
46. Ito Y, Corna S, vonBrevern M, Bronstein A, Gresty H. The functional effectiveness of neck muscle reflexes for head-righting in response to sudden fall. Exp Brain Res. 1997;117(2):266–72. https://doi.org/10.1007/s002210050221.
47. Kuramochi R, Kimura T, Nakazawa K, Akai M, Torii S, Suzuki S. Anticipatory modulation of neck muscle reflex responses induced by mechanical perturbations of the human forehead. Neurosci Lett. 2004;366(2):206–10. https://doi.org/10.1016/j.neulet.2004.05.040.
48. Portero R, Quaine F, Cahouet V, Thoumie P, Portero P. Musculo-tendinous stiffness of head–neck segment in the sagittal plane: An optimization approach for modeling the cervical spine as a single-joint system. J Biomech. 2013;46(5):925–30. https://doi.org/10.1016/j.jbiomech.2012.12.009.
49. Portero R, Lecompte J, Thoumie P, Portero P. Musculo-tendinous stiffness of the in vivo head-neck segment in response to quick-releases: a reproducibility study. Isokinet Exerc Sci. 2011;19(1):7–12.
50. Tierney RT, Sitler MR, Swanik CB, Swanik KA, Higgins M, Torg J. Gender differences in head-neck segment dynamic stabilization during head acceleration. Med Sci Sports Exerc. 2005;37(2):272–9. https://doi.org/10.1249/01.mss.0000152734.47516.aa.
51. Alsalaheen B, Bean R, Almeida A, Eckner J, Lorincz M. Characterization of cervical neuromuscular response to head-neck perturbation in active young adults. J Electromyogr Kinesiol. 2018;39:70–6. https://doi.org/10.1016/j.jelekin.2018.01.011.
52. Portero R, Quaine F, Cahouet V, Lecompte J, Thoumie P, Portero P. In vivo neck musculo-tendinous stiffness in response to quick-releases. In: 6th world congress of biomechanics (Wcb 2010), Pts 1-32010, pp. 593–6.
53. Vasavada A, Trask D, Knottnerus A, Lin D (eds). Effects of head position and impact direction on neck muscle response to perturbations. In: American society of biomechanics proceedings; 2009.
54. Debison-Larabie C. Examining the relationship between cervical anthropometrics, head kinematics and cervical muscle responses to sudden head perturbations in competitive ice hockey players. Oshawa: University of Ontario Institute of Technology; 2016.
55. Lucas GQ. A mechanical apparatus to quantify the reflex response of the human head/neck system. Pullman: Washington State University; 2006.
56. Rezasoltani A, Ylinen J, Bakhtiary A, Norozi M, Montazeri M. Cervical muscle strength measurement is dependent on the location of thoracic support. Br J Sports Med. 2008;42(5):379–82.
57. Matheson L, Mooney V, Caiozzo V, Jarvis G, Pottinger J, DeBerry C, et al. Effect of instructions on isokinetic trunk strength testing variability, reliability, absolute value, and predictive validity. Spine. 1992;17(8):914–21.
58. Mihalik JP, Blackburn JT, Greenwald RM, Cantu RC, Marshall SW, Guskiewicz KM. Collision type and player anticipation affect head impact severity among youth ice hockey players. Pediatrics. 2010;125(6):e1394–401.
59. Siegmund GP, Blouin J-S, Inglis JT. Does startle explain the exaggerated first response to a transient perturbation? Exerc Sport Sci Rev. 2008;36(2):76–82. https://doi.org/10.1097/JES.0b013e318168f1ce.
60. Siegmund GP, Sanderson DJ, Myers BS, Inglis JT. Rapid neck muscle adaptation alters the head kinematics of aware and unaware subjects undergoing multiple whiplash-like perturbations. J Biomech. 2003;36(4):473–82.
61. Nijhuis LBO, Allum JH, Borm GF, Honegger F, Overeem S, Bloem BR. Directional sensitivity of “first trial” reactions in human balance control. J Neurophysiol. 2009;101(6):2802–14.
62. Tucker R, Raftery M, Kemp S, Brown J, Fuller G, Hester B, et al. Risk factors for head injury events in professional rugby union: a video analysis of 464 head injury events to inform proposed injury prevention strategies. Br J Sports Med. 2017;51(15):1152–7. https://doi.org/10.1136/bjsports-2017-097895.
63. Salmon DM, Handcock P, Niven B. Can neck strength be measured using a single maximal contraction in a simulated contact position? J Strength Cond Res. 2018;32(8):2166–73. https://doi.org/10.1519/JSC.0000000000001874.
64. Gilchrist I, Moglo K, Storr M, Pelland L. Effects of head flexion posture on the multidirectional static force capacity of the neck. Clin Biomech. 2016;37:44–52.
65. Hildenbrand KJ, Vasavada AN. Collegiate and high school athlete neck strength in neutral and rotated postures. J Strength Cond Res. 2013;27(11):3173–82.
66. Penning L. Normal movements of the cervical spine. Am J Roentgenol. 1978;130(2):317–26.
67. Guskiewicz KM, Mihalik JP, Shankar V, Marshall SW, Crowell DH, Oliaro SM, et al. Measurement of head impacts in collegiate football players: relationship between head impact biomechanics and acute clinical outcome after concussion. Neurosurgery. 2007;61(6):1244–52. https://doi.org/10.1227/01.neu.0000306103.68635.1a **(discussion 52–3)**.
68. Weik MH. Nyquist theorem. In: Computer science and communications dictionary. Boston: Springer; 2001. p. 1127.
69. Wu LC, Laksari K, Kuo C, Luck JF, Kleiven S, ‘Dale’ Bass CR, et al. Bandwidth and sample rate requirements for wearable head impact sensors. J Biomech. 2016;49(13):2918–24. https://doi.org/10.1016/j.jbiomech.2016.07.004.
70. Rowson S, Duma SM, Beckwith JG, Chu JJ, Greenwald RM, Crisco JJ, et al. Rotational head kinematics in football impacts: an injury risk function for concussion. Ann Biomed Eng. 2012;40(1):1–13. https://doi.org/10.1007/s10439-011-0392-4.
71. Guskiewicz KM, Mihalik JP. Biomechanics of sport concussion: quest for the elusive injury threshold. Exerc Sport Sci Rev. 2011;39(1):4–11. https://doi.org/10.1097/JES.0b013e318201f53e.
72. Zhang L, Yang KH, King AI. A proposed injury threshold for mild traumatic brain injury. J Biomech Eng. 2004;126(2):226–36.

73. Elkin BS, Elliott JM, Siegmund GP. Whiplash injury or concussion? A possible biomechanical explanation for concussion symptoms in some individuals following a rear-end collision. J Orthop Sports Phys Ther. 2016;46(10):874–85. https://doi.org/10.2519/jospt.2016.7049.
74. Gadd CW. Use of a weighted-impulse criterion for estimating injury hazard: SAE technical paper 1966. Report no. 0148-7191.
75. Greenwald RM, Gwin JT, Chu JJ, Crisco JJ. Head impact severity measures for evaluating mild traumatic brain injury risk exposure. Neurosurgery. 2008;62(4):789–98. https://doi.org/10.1227/01.neu.0000318162.67472.ad **(discussion 98)**.
76. Brennan JH, Mitra B, Synnot A, McKenzie J, Willmott C, McIntosh AS, et al. Accelerometers for the assessment of concussion in male athletes: a systematic review and meta-analysis. Sports Med. 2016;47(3):469–78.
77. Patton DA. A review of instrumented equipment to investigate head impacts in sport. Appl Bionics Biomech. 2016. https://doi.org/10.1155/2016/7049743 **(Article ID 7049743)**.
78. Patricios J, Fuller GW, Ellenbogen R, Herring S, Kutcher JS, Loosemore M, et al. What are the critical elements of sideline screening that can be used to establish the diagnosis of concussion? A systematic review. Br J Sports Med. 2017;51(11):888–94.
79. King D, Hume P, Gissane C, Brughelli M, Clark T. The influence of head impact threshold for reporting data in contact and collision sports: systematic review and original data analysis. Sports Med. 2016;46(2):151–69. https://doi.org/10.1007/s40279-015-0423-7.
80. Tierney GJ, Lawler J, Denvir K, McQuilkin K, Simms CK. Risks associated with significant head impact events in elite rugby union. Brain Inj. 2016;30(11):1350–61. https://doi.org/10.1080/02699052.2016.1193630.
81. Kemp SP, Hudson Z, Brooks JH, Fuller CW. The epidemiology of head injuries in English professional rugby union. Clin J Sport Med. 2008;18(3):227–34.
82. Emery CA, Kang J, Shrier I, Goulet C, Hagel BE, Benson BW, et al. Risk of injury associated with body checking among youth ice hockey players. JAMA. 2010;303(22):2265–72.
83. Shewchenko N, Withnall C, Keown M, Gittens R, Dvorak J. Heading in football. Part 1: Development of biomechanical methods to investigate head response. Br J Sports Med. 2005;39(Suppl. 1):i10–25. https://doi.org/10.1136/bjsm.2005.019034.
84. Pellman EJ, Viano DC, Tucker AM, Casson IR. Concussion in professional football: Location and direction of helmet impacts—part 2. Neurosurgery. 2003;53(6):1328–41.
85. Gennarelli TA, Thibault LE, Adams JH, Graham DI, Thompson CJ, Marcincin RP. Diffuse axonal injury and traumatic coma in the primate. Ann Neurol. 1982;12(6):564–74.
86. Patton DA, McIntosh AS, Kleiven S. The biomechanical determinants of concussion: finite element simulations to investigate brain tissue deformations during sporting impacts to the unprotected head. J Appl Biomech. 2013;29(6):721–30.
87. Kleiven S. Predictors for traumatic brain injuries evaluated through accident reconstructions. Stapp Car Crash J. 2007;51:81.
88. Dick RW. Is there a gender difference in concussion incidence and outcomes? Br J Sports Med. 2009;43(Suppl 1):i46–50. https://doi.org/10.1136/bjsm.2009.058172.