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Effect of jaw clenching on head acceleration during a predictable load impact

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Abstract

Background: Jaw clenching is considered to reduce head acceleration while receiving a strong impact on the body during sport activities.

Objective: The present study aimed to clarify the effect of jaw clenching on reduction of head acceleration during a predictable load impact to the body.

Methods: Seven healthy participants were exposed to a predictable load impact with and without jaw clenching. We recorded the electromyographic activity of the masseter (MA) and digastricus (DIG) muscles, occlusal pressure and head acceleration throughout the experiment.

Results: When participants were not instructed to clench their jaws, they naturally positioned their jaws without occlusal contact at the time of pendulum impact by co-contracting the jaw opener and closer muscles. When participants were instructed to clench their jaws, neither the activity of the jaw opener muscle nor the head acceleration differed at the time of pendulum impact when compared with when participants were not instructed to clench their jaws.

Conclusions: A slightly distanced jaw position (co-contracting the jaw opener and closer muscles without occlusal contact) might serve inherently safety for reduction of head acceleration during predictable body impact, while jaw clenching does not contribute to reduction of head acceleration in response to pendulum impact more than the distanced jaw position does. Notably, DIG activation to minimise the head acceleration in response to pendulum impact was similar in clenching and no clenching positions. This suggests that DIG may play a crucial role in the reduction of head acceleration, regardless of MA muscle activity.

KEYWORDS

acceleration, athletes, jaw, mandible, masseter muscles

1 | INTRODUCTION

Jaw clenching is thought to reduce head acceleration while receiving a strong impact on the body during sport activities.[1,2] For example, a previous study conducted among athletes reported that head acceleration during a tackle was less when rugby players clenched their jaws than when they did not.[1] Similarly, in soccer, head acceleration during heading was less when the players clenched their jaws than when they did not.[2] These studies found that the activity of the neck muscles increased with an increase in jaw clenching, resulting in a decrease in head acceleration in response to the impact. It was reported that strengthened neck muscles are associated with a lower risk of concussion in an observational study conducted among 6700 adolescent contact sport athletes[3] and a laboratory study conducted among contact sport athletes of varying ages.[4,5]

In contrast, we found that there were sport activities during which occlusal contact was not seen, which investigated athletes' occlusal contact (i.e., contact of the upper and lower teeth) and activities of the jaw closer and opener muscles during sports.[6,7] For example, in Judo, occlusal contacts were not observed even in response to strong body perturbations, such as being thrown down, during matches, though considerable activity was observed in both jaw closer and opener muscles.[6] Similarly, in our previous study conducted among 10 boxers (all with more than three years of experience), the occlusal contact in all participants was less than 5% of the total occlusal surface area throughout the matches, including when receiving punches.[7] Furthermore, previous studies have reported that during strong skeletal muscle force exertion in dynamic exercises, such as deadlift, the jaw opener and closer muscles simultaneously contract to position the mandible so as not to have occlusal contact between upper and lower teeth, distinct from the maximal intercuspal position (i.e., in which all teeth contact simultaneously).[8] Similar results were found during deadlifting that requires effective execution of power stroke of muscle contraction.[8] Thus, these studies suggest that athletes may naturally keep their jaw positions apart from each other with a distance co-contraction of jaw opener and closer muscle to prevent traumatic collision of upper and lower teeth during the power stroke of body muscle contractions.

The monitoring of masseter (MA) muscle activity alone as a representative indicator of jaw clenching is not necessarily sufficient, since there are vital differences between jaw clenching in the intercuspal position (i.e., the occlusal position where all of the teeth contact simultaneously) and other jaw positions (i.e., in which the co-contraction of jaw opener and closer muscles fixes participants' jaws in a distanced position without occlusal contact). Simultaneous measurement of occlusal contact and the activity of jaw closer and opener muscles are necessary to differentiate these two jaw positions. The present study used a system that measures occlusal contact and activity of jaw closer and opener muscles continuously and aimed to clarify the effect of jaw clenching on reduction of head acceleration during predictable body impacts.

2 | MATERIALS AND METHODS

2.1 | Participants

The participants in this study were university students who were recruited by posting flyers around the university. Seven fully dentate healthy young participants (five males and two females with a mean age of 22.0 ± 0.5 years, a mean height of 168.7 ± 6.6 cm and a mean body weight 61.5 ± 16.7 kg) meeting the following criteria were enrolled in this study. The following inclusion criteria were applied: the presence of a Class I incisor relationship (i.e., the lower incisor edges occlude with or lie immediately below the cingulum plateau of upper central incisors), the absence of mobile teeth (Miller Classification II or III, i.e., >1 mm horizontal and/or vertical mobility),[9] the absence of subjective temporomandibular problems[10] and the absence of musculoskeletal or neurological pathology that could affect task performance. This study was approved by the ethics committee of Osaka Dental University (approval number: 111088), and all participants provided their written informed consent prior to study participation. This study was conducted in accordance with the provisions of the Declaration of Helsinki as well as the published ethical standards of this journal.[11]

2.2 | Experimental set-ups

The participants were exposed to impact induced by a pendulum device, according to the methodology of previous studies (Figure 1A).[12–14] Participants stood barefoot on a force platform (BP400600, AMTI, MA, USA) with their feet placed shoulder-width apart (Figure 1A). The foot position was marked on the floor to maintain participants' body location relative to the perturbation device throughout the duration of the experiment. A load (mass: m = 3% of each participant's body weight) was attached to the distal end of the pendulum, and its initial height was adjusted to each participant's height (height: h = 150% of participant height) (Figure 1A). An accelerometer (myoMOTION, Noraxon, sampling rate: 200 Hz) was attached to the right dorsal hand to identify the moment of impact (Figure 1A).

The electromyographic (EMG) activities of the MA and digastricus (DIG) muscles were recorded at 2000 Hz (Ultium, Noraxon). After the skin area was cleaned with alcohol preps, disposable surface electrodes (G207, Nihon Kohden) were attached to the muscle belly of the MA and DIG. A ground electrode was embedded in each EMG electrode. The distance between each pair of electrodes was 20 mm. The muscles on the right side of the body were evaluated.

Occlusal contact was monitored using the I-Scan system that measures pressure distribution (I-Scan, Nitta Corporation, sampling rate: 100 Hz). The system is composed of a pressure sensor sheet (FPD-T-Sports, Nitta Corporation), data acquisition electronics, and a personal computer. The pressure sensor sheet consists of a *U*-shaped sensing area and a tab that connects to data acquisition electronics. The tab of the pressure sensor sheet was extended to a length that did not disturb participants' activities during measurement. The *U*-shaped sensing area of the pressure sensor sheet was attached to the upper dental arch with an oral apparatus (Figure 1B), as detailed below. The system has been scrutinised in various studies that have alternately supported[15–18] and contradicted[19,20] its accuracy. Additionally, though occlusal force is a vector with a magnitude and a direction, the direction of the force could not be assessed since the system can only provide a relative magnitude. However, this system is superior in terms of verifying participants' constant occlusal contact than any other methods.

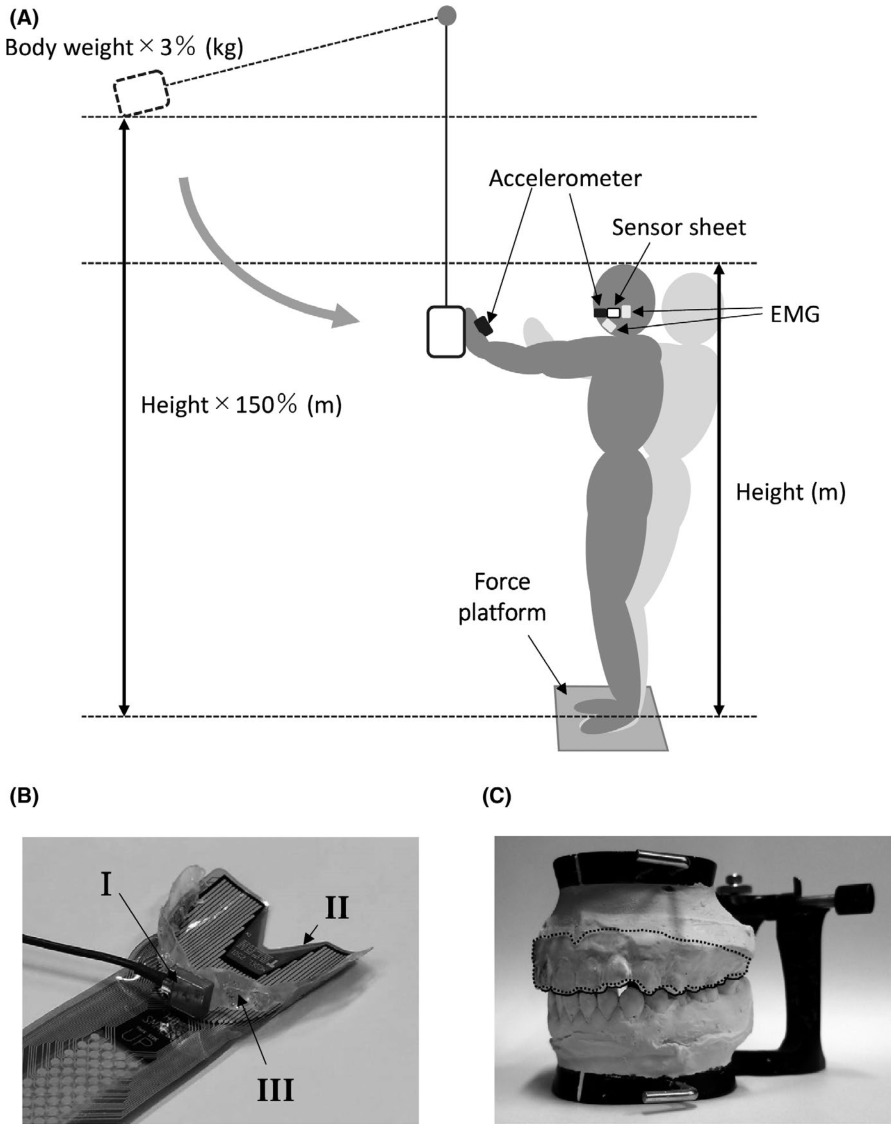

FIGURE 1 (A) Schema of the experimental setup. Participants were exposed to impact induced using a pendulum device. A load was attached to the distal end of the pendulum, and its initial height was adjusted to the participants' height. Surface electrodes were attached to the digastricus and masseter muscles. A pressure sensor sheet measuring occlusal pressure was attached to the upper dental arch, and an accelerometer was attached to the upper anterior tooth and the right hand; (B) the following sensors were affixed to the participants' upper dental arch: (I) an accelerometer measuring head acceleration, (II) a pressure sensor sheet measuring occlusal pressure and (III) an oral apparatus fixing the sensors to the upper dental arch; and (C) articulated working models fabricating a custom-made oral apparatus for fixing the pressure sensor sheet and the accelerometer to the upper dental arch. The dotted line represents the outline of the oral apparatus, 1 mm away from the cusp line (solid line) of the upper dental arch, so that the oral apparatus does not interfere with participants' occlusion

An accelerometer (AMA-A, Kyowa Dengyo, sampling rate: 500 Hz) was attached to the upper anterior tooth region (a hard tissue directly connected to the head) with an oral apparatus to measure head acceleration during impact. A custom-made oral apparatus was fabricated to attach the sensor sheet and accelerometer to the participants' teeth. An oral apparatus for the maxillary arch was fabricated through the following procedures for each participant. Dental impressions (Aroma Fine Plus, GC, Tokyo, Japan) were taken, and a 3-mm-thick polyester sheet (DURAN Plus, JM Ortho, Tokyo, Japan) was pressed on the working models (New Plastone, GC, Tokyo, Japan) using a pressure-forming machine (MINISTAR S, J. MORITA CORP). Working models were articulated to fabricate a custom-made oral apparatus for attaching the pressure sensor sheet and the accelerometer to the participants' upper dental arch. The oral apparatus was fabricated to fit the buccal side of the upper dental arch of each participant. Its outline was set 1 mm away from the cusp line of the upper dental arch (the solid line in Figure 1C), so that the oral apparatus does not interfere with the participants' occlusion. The pressure sensor sheet and the accelerometer were attached to the oral apparatus using instant adhesive (Aron Alpha, Daiichi Sankyo) (Figure 1B). The oral apparatus was fixed with the pressure sensor sheet, and the accelerometer was adhered to the upper dental arch of each participant using a denture adhesive (Poligrip S, Glaxo Smith Kline Consumer Healthcare Japan, Tokyo, Japan). Proper retention, which removed the possibility of sensors detaching when the lips and cheeks moved, was checked intraorally by a dentist licenced to practice prosthodontics and sport dentistry.

Electromyography and acceleration signals were measured through an integrated data recording system (myoMUSCLE, Noraxon). Analog synch signals were used to synchronise EMG, acceleration and I-Scan signals through a sensor interface (VICON MX, Oxford Metrics). The centre-of-pressure position was automatically calculated from the moments and force measured using the force platform, and this position was displayed online. We used the displayed centre-of-pressure position for monitoring the initial centre-of-pressure position.

2.3 | Experimental procedures

Participants were required to receive the pendulum impact with their hands, while their arms, wrists and fingers were extended at the shoulder level (Figure 1A). The initial centre-of-pressure position was monitored by the research assistant to ensure consistency of the experimental conditions. Participants were asked to maintain their balance after the impact without taking a step. The load attached to the distal end of the pendulum was released towards the participants in a sagittal plane after a countdown to allow the participants to predict the timing of the load impact. Prior to the experiment, maximal voluntary contraction (MVC) of the MA and DIG was measured to acquire 100% MVC for EMG data. We also measured the maximum occlusal pressure during maximum voluntary jaw clenching at the intercuspal position. The participants were exposed to the impact under no clenching and clenching conditions. Under the no clenching condition, participants were asked not to clench their jaw prior to each single impact to clarify effect of jaw clenching on head acceleration during a predictable load impact. Under the clenching condition, participants were instructed to maintain jaw clenching at a 30% MVC level at the intercuspal position throughout the recording. To avoid fatigue of participants' jaw muscles, we stopped the recording 3 s after the impact. A research assistant monitored the participants' jaw clenching and occlusal contact and provided verbal feedback to the participant if these deviated from target activation levels. The countdown for the release of the pendulum was started once the clenching level reached its target (i.e., 30% MVC). In total, 20 trials were recorded, where the participants received 10 consecutive predictable impacts under the clenching condition and another 10 consecutive impacts under the no clenching condition. The order of the experimental conditions was allocated randomly, using the rand function in Excel (Microsoft). Two to three practice trials were performed prior to initial testing. An interval of 30–45 s was set between the trials to avoid fatigue.

2.4 | Data processing

The data were analysed using a custom-made MATLAB program (MATLAB 2014b, MathWorks). Time-zero (T_0; the moment of the impact) was defined as the time at which the tangential acceleration of the hand, which was acquired using the accelerometer, reached 5% of its peak value.

The EMG signals were downsampled to 1000 Hz using the resample function of MATLAB. Resampled EMG signals were rectified and filtered with the Butterworth band-pass filter (2nd order, 50–500 Hz). We calculated moving averages for the filtered EMG signals based on a 50-ms time window. The EMG activities of DIG and MA were normalised by 100% MVC for each participant. The onset times of DIG and MA activity were defined as the time at which normalised EMG activity became greater than background normalised EMG activities (mean ± 2SD within the time window from T_0 −1,000 ms to T_0 - 500 ms) for more than 50 ms within the time window from T_0 - 250 ms to T_0 + 500 ms. The peak values of DIG and MA activity were defined as the maximal values within the time window from T_0 - 250 ms to T_0 + 500 ms. The identification of the onset time and the peak value was performed by combining computer algorithms with visual inspection of the trials. We also calculated the increase in EMG activity at its peak. The increase in EMG activity at its peak was defined as the value that subtracted a baseline from the peak value. There was no significant difference in the baseline of DIG activity between the two conditions (Mean ± SE: no clenching: 0.05 ± 0.01, clenching: 0.05 ± 0.01, p = .35), unlike the baselines of MA activity that significantly differed (no clenching: 0.02 ± 0.004, clenching: 0.28 ± 0.01, p < .05).

The tangential head acceleration signals, which were calculated from the three axial accelerations at each timepoint, were filtered with the Butterworth low-pass filter (2nd order, 20 Hz). Peak values and onset times of the tangential head accelerations were identified. First and second peak values were defined as the maximal and second largest values within the time window from T_0 to T_0 + 500 ms, respectively. Onset time was defined as the time at which the totalled head acceleration reached 5% of first peak value within the time window from T_0 – 250 ms to T_0 + 500 ms.

2.5 | Statistical analysis

The mean values of each condition for each participant were used for the analysis. The Wilcoxon signed-rank test was performed to compare differences between muscles or conditions. All group data are represented as medians ± interquartile ranges (IQR). As for the EMG activity, the following parameters were compared between muscles (i.e., DIG and MA) or conditions (i.e., no clenching and clenching conditions): the onset time (closed triangles in Figure 2A), time-to-peak and the increase in EMG activity at its peak (Figure 2A). As for the head acceleration, the following parameters were compared between conditions (i.e., no clenching and clenching conditions): the onset time, peak value and time-to-peak (Figure 2C). Statistical analysis was performed using EZR version 3.4.1 (Jichi Medical University, Saitama, Japan) at a 5% significance level.[21] We used G*power 3.1 (Heinrich-Heine-Universität, Düsseldorf, Germany) to estimate the sample size required using Wilcoxon's signed-rank test.[22,23] The effect size (r), which was used to calculate the sample size, was derived from the mean and standard deviation (SD) reported in a previous study that compared head acceleration between no clenching and clenching conditions during impact in sports.[2] The study reported that the means ± SDs of head acceleration were 28.4 ± 7.0 G and 23.9 ± 6.2 G for the no clenching and the clenching conditions, respectively. A total sample of seven was required for an alpha probability of 0.05 and a statistical power of 0.80.

3 | RESULTS

All participants completed the study tasks without any difficulty. No adverse effects were reported after participation in the study.

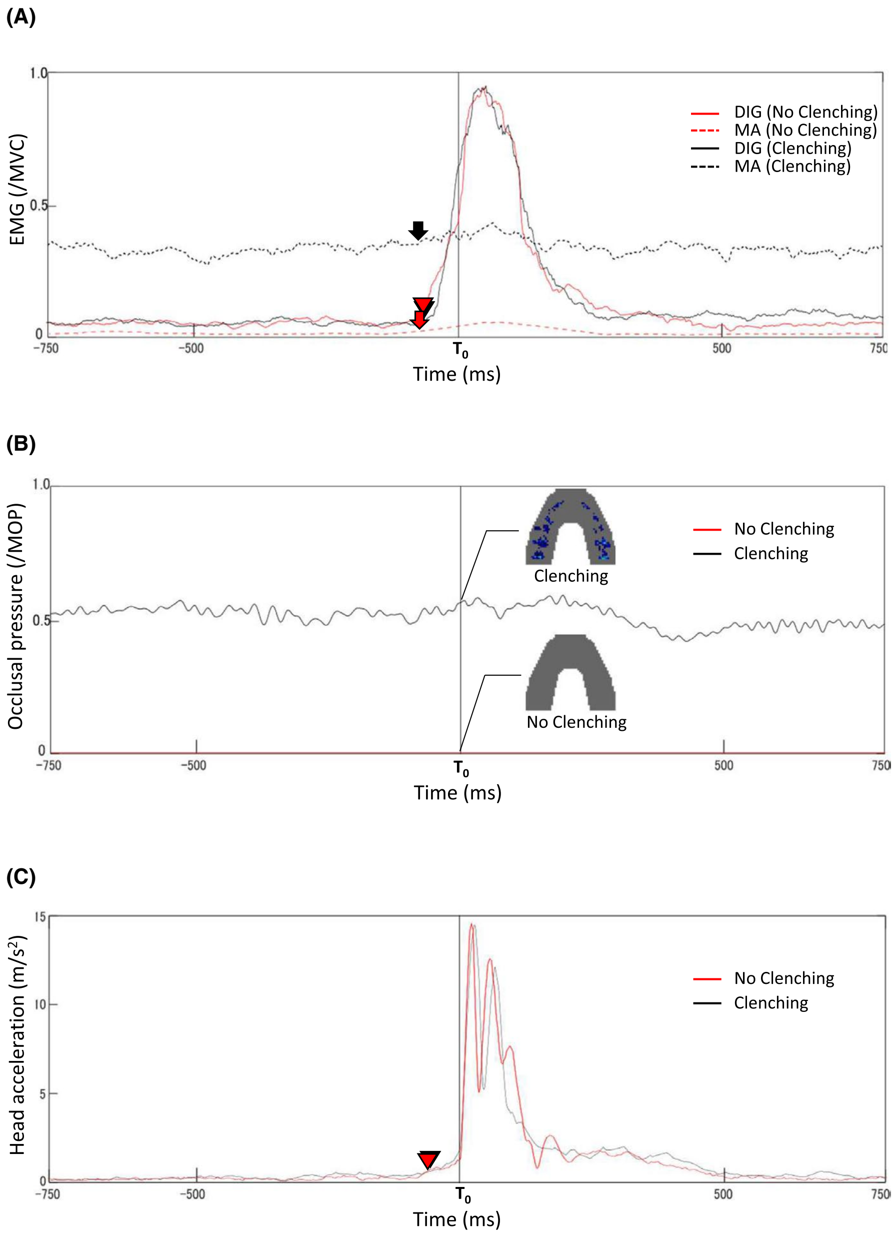

FIGURE 2 Representative examples of waveforms (T0: moment of impact). The red line indicates the data obtained under the no clenching condition, whereas the black line indicates data obtained under the clenching condition. (A) Rectified and filtered electromyographic (EMG) signals are shown. The solid line indicates the EMG signal of the digastricus muscles (DIG). The dotted line indicates the EMG signal of the masseter muscles (MA). Each signal was normalised by each maximum voluntary contraction (MVC). The closed red and black triangles indicate the onset of the DIG activity under the no clenching and clenching conditions, respectively. The red and black arrowheads indicate the onset of the MA activity under the no clenching and clenching conditions, respectively. The onset times of DIG and the MA activity were defined as the times at which EMG activity became greater than background EMG activity for more than 50 ms within the time window from T_0 − 250 ms to T_0 + 500 ms. (B) The occlusal pressure signals and the pressure distribution at the time of the impact are shown. The occlusal pressure signal was normalised by the maximum occlusal pressure (MOP). The occlusal pressure was absent under the no clenching condition, while pressure distribution was more even across the tooth arch under the clenching condition. (C) Rectified and filtered head acceleration signals are shown. The peak value of the head acceleration was defined as the maximal value from T0 − 250 ms to T0 + 500 ms. The black and red triangles indicate the onset of the head acceleration under the no clenching and clenching conditions, respectively. The onset of head acceleration was defined as the time at which acceleration became greater than 10% of its peak for more than 50 ms

Representative examples of waveforms of EMG activity for MA and DIG, occlusal pressure and head acceleration observed in response to the pendulum impact are shown in Figure 2A–C, respectively.

3.1 | DIG and MA activity

For the between-muscle comparison in the no clenching condition, EMG activity of both DIG and MA started to simultaneously increase,

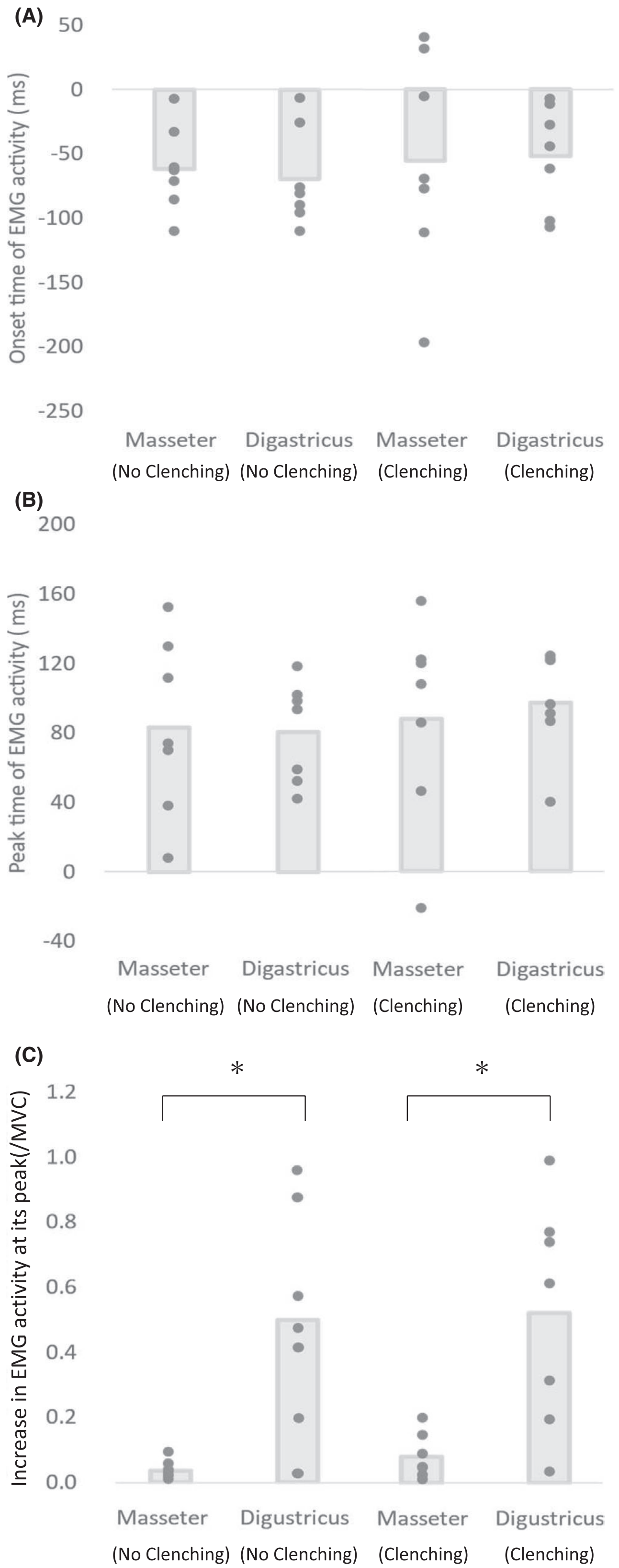

FIGURE 3 The comparison of pooled data for the electromyographic (EMG) activity between muscles (digastricus and masseter muscles) and conditions (no clenching and clenching conditions). The mean values in the no clenching and clenching conditions, for the seven participants, are presented in the bar graph. (A) The onset time. (B) The peak time. (C) The increase in EMG activity at its peak. $^*p < .05$. MVC: maximum voluntary contraction

slightly prior to T_0 (DIG: −80.96 ± 41.72 ms; MA: −63.16 ± 31.56 ms, $p = .38$, Figure 3A). The time-to-peak EMG activity did not differ between the two muscles (DIG: 93.30 ± 40.49 ms; MA: 73.80 ± 66.56 ms, $p = .94$; Figure 3B). In contrast, the increase in DIG activity at its peak was greater than that of MA (DIG: 0.50 ± 0.42; MA: 0.04 ± 0.03, $p = .02$; Figure 3C). Similarly, under clenching condition, there was no significant difference in onset time (DIG: −51.60 ± 62.58 ms; MA: −55.32 ± 107.27 ms, $p = .81$; Figure 3A) and time-to-peak (DIG: 97.44 ± 33.32 ms; MA: 88.05 ± 55.01 ms, $p = .69$; Figure 3B), whereas the increase in DIG activity at its peak was greater than that of MA (DIG: 0.52 ± 0.50; MA: 0.08 ± 0.12, $p = .05$; Figure 3C).

For the between-condition comparison in DIG, there was no significant difference in (1) EMG onset time (no clenching: −80.96 ± 41.72 ms; clenching: −44.44 ± 62.58 ms, $p = .30$; Figure 3A), (2) time-to-peak EMG (no clenching: 93.30 ± 44.49 ms; clenching: 96.30 ± 33.32 ms, $p = .11$; Figure 3B) and (3) increase in EMG activity at its peak (no clenching: 0.50 ± 0.42; clenching: 0.52 ± 0.50, $p = .47$; Figure 3C). Similarly, the MA activity showed no significant differences in (1) onset time (no clenching: −61.91 ± 55.32 ms; clenching: −44.44 ± 62.58 ms, $p = .81$; Figure 3A), (2) time-to-peak EMG (no clenching: 83.17 ± 66.56 ms; clenching: 88.05 ± 55.01 ms, $p = .30$; Figure 3B) and (3) increase in EMG activity at its peak (no clenching: 0.04 ± 0.03; clenching: 0.08 ± 0.12, $p = .16$; Figure 3C) between the no clenching and clenching conditions.

3.2 | Occlusal pressure

Occlusal pressure was not observed during the experiment in the no clenching condition, whereas it was maintained at approximately 50% under the clenching condition (clenching: 0.50 ± 0.24; Figure 4). Similarly, the occlusal contact was absent under the no clenching condition, while all teeth contact simultaneously under the clenching condition.

3.3 | Head acceleration

Head acceleration occurred slightly after T_0 and rapidly increased to the peak, was followed by a second peak and then finally returned to the previous level under both the no clenching and clenching conditions. There was no significant difference in (1) acceleration onset time (no clenching: −130.80 ± 12.08 ms; clenching: −128.25 ± 17.27 ms, $p = .38$; Figure 5A), (2) first peak acceleration (no clenching: 13.68 ± 3.08 m/s^2; clenching: 13.89 ± 4.56 m/s^2, $p = .69$; Figure 5B), (3) time-to-first peak (no clenching: 37.14 ± 7.18 ms; clenching: 41.20 ± 6.37 ms, $p = .81$; Figure 5C) and (4) second peak value (no clenching: 7.89 ± 3.79 ms; clenching: 7.70 ± 3.96 ms, $p = .08$; Figure 5D) between the conditions.

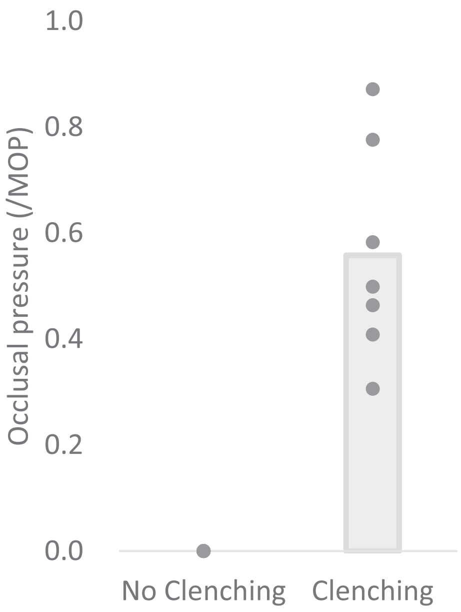

FIGURE 4 The pooled data for the occlusal pressure. The mean values in the no clenching and clenching conditions, for the seven participants, are presented in the bar graph. MOP, maximum occlusal pressure

4 | DISCUSSION

This study aimed to clarify the effect of jaw clenching on protecting the head from traumatic damage to the body. We observed three main findings. First, under the no clenching condition, the activity of the jaw opener and closer muscles started increasing simultaneously, slightly prior to the impact; peaked; and subsequently returned to the baseline after the impact. In addition, no participant had occlusal contact throughout the experiment under the no clenching condition. Second, there was no significant difference in onset time, time-to-peak of the EMG activities or the increase in EMG activity at its peak between conditions (with and without jaw clenching). Third, between clenching and no clenching conditions, there was no significant difference in the first and second peak values of head acceleration induced in response to the same pendulum load impact.

The mandible should be fixed in a position that is optimal for the prevention of a traumatic impact to the body. Our results suggest that participants instinctively fixed their jaws in a slightly distanced position by co-contracting the jaw opener and closer muscles without occlusal contact against impact in the no clenching condition. Other previous studies have postulated that to keep participants' jaws stable during rapid head movement could be inherently safe in terms of eliminating hazardous trauma.[6–8] We previously found that occlusal contact was almost absent when athletes experienced impact during martial arts sporting activities, such as judo, boxing and nippon kenpo.[6,7,24] From a physiological point of view, the stabilisation of the jaw position during locomotion is controlled by reflex pathways. One reflex (the stretch reflex) contributes to preventing the jaw from bouncing during locomotion by maintaining a fixed jaw position.[25] Another reflex (the unloading reflex) is responsible for the prevention of collision of the upper and lower teeth.[26] This suggests that jaw fixation without occlusal contact may play a beneficial role in the prevention of trauma. It is noteworthy that there was a significant difference in the increase in EMG activity at its peak between DIG and MA at the time of co-contraction as shown in Figures 2A and 3C. A previous study conducted among 26 participants revealed that the maximal forces which can be produced by knee and elbow muscles are proportional to their cross-sectional area.[27] Then, because the EMG activity during MVC reflects the cross-sectional area of the muscle which is much smaller in DIG than in MA, the DIG needs to be activated much more than the MA to achieve the fixed position by co-contraction when compared as the EMG activity normalised to its MVC.

Our results indicate that jaw clenching at the intercuspal position (i.e., the occlusal position where all teeth contact simultaneously) affected neither the head acceleration nor the activity of the jaw opener muscle during the impact to the body, though several studies have reported regarding the effectiveness of jaw clenching for the reduction of head acceleration.[1,2] Such studies should be interpreted with caution because of the following two reasons. First, those studies could not accurately confirm that participants clench their jaws during the test exercise and may have overlooked an unclenching of the teeth. This was because these studies only evaluated the activity of the MA to monitor jaw clenching, which does not differentiate jaw clenching at the intercuspal position from jaw fixation in a slightly distanced position (i.e., a jaw position where upper and lower teeth are slightly separated while co-contractions of the jaw opener and closer muscles occur). We overcame this limitation by monitoring the occlusal contacts with a sensor sheet placed between the upper and lower teeth, in addition to monitoring the activity of the MA. This sensor sheet enabled direct and spontaneous monitoring of the occlusal contact throughout the experiment. Second, head acceleration is difficult to measure accurately. In most previous studies, accelerometers were attached to a helmet or a headband. Therefore, there was a possible cushioning effect from inclusions (such as helmets or headgear) between the skull and accelerometers, potentially resulting in inaccurate measurement of head accelerations. Accelerometers mounted directly into or on helmets have previously been shown to overestimate head motion,[28,29] and helmet fit can affect the accuracy of the system.[30] We overcame these limitations by attaching the accelerometer to the upper teeth, a hard tissue directly connected to the head skull, to minimise the risk of overestimating head acceleration. As found in this study, DIG activity largely increased in response to pendulum impact even during clenching. Nevertheless, occlusal pressure remained almost constant because MA and DIG were simultaneously activated under the clenching and no clenching conditions in response to the impact (Figure 2A). Thus, unclenching was not induced in response to the impact.

Contrastingly, some epidemiological studies have reported that the use of a mouthguard is effective for the prevention of sport-related concussions.[31,32] These results could be explained by the effectiveness of mouthguards with respect to ensuring a fixed jaw

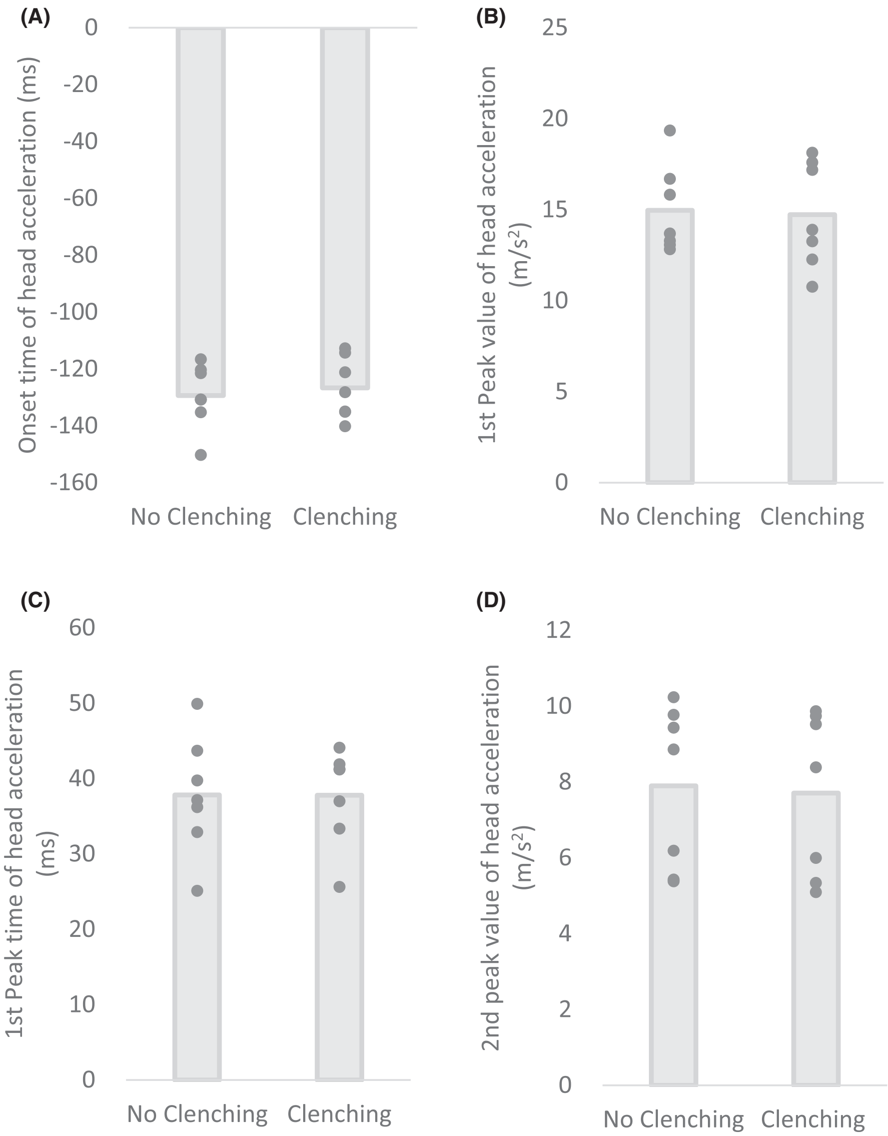

FIGURE 5 The comparison of pooled data for the head acceleration activities in the no clenching and clenching conditions. The mean values in the no clenching and clenching conditions, for the seven participants, are presented in the bar graph. (A) The onset time of head acceleration ($p = .38$). (B) The first peak value of head acceleration ($p = .69$). (C) The first peak time of head acceleration ($p = .81$). (D) The second peak value of head acceleration ($p = .08$)

position. For example, several studies have revealed that the use of a mouthguard increases the activity of neck muscles, which may assist in fixing the jaw, which then decreases head acceleration during impact.[33,34] Further studies are required to elucidate how the use of a mouthguard decreases the risk of concussions.

Our results should be interpreted with caution because of following four reasons. First, this study was conducted on a small sample size, which may have caused type II errors (i.e., false negatives) during statistical analyses. However, clinically meaningful differences were not observed qualitatively for comparisons of $p > .05$. For example, the difference in mean head acceleration between the no clenching and clenching conditions was 0.2 m/s^2 (a 1.3% difference). Therefore, we believe that type II errors were avoided. Second, the impact that participants experienced was much smaller than an impact that would cause a concussion. However, this limitation cannot be overcome without violating ethical standards. A computer simulation is necessary to reproduce the impact that would cause a concussion. Furthermore, a computer cannot easily simulate the characteristics of multiple tissues (such as the bone, muscle and skin) that possess various biomechanical properties or various physical functions, such as postural control to compensate for the impact. Nonetheless, we believe that the results of this study provide beneficial data among human participants. Third, in the previous studies,[1,2] impacts were received with the chest or head, but not with the hand of an extended arm. Direct impact to the body trunk may have more severe traumatic effects than that to the hands. Lastly, findings of the present study are limited to an external perturbation applied in the sagittal plane. Studies have shown that the direction of the perturbation modulates muscle activities of trunk and leg muscles.

In conclusion, our results suggest that the participants naturally fixed their jaws in a slightly distanced position by co-contracting the jaw opener and closer muscles without occlusal contact against impact. Our results further indicated that jaw clenching at the intercuspal position did not affect either the activity of the jaw opener muscle or head acceleration during the impact to the body. On the contrary, we believe that a slightly distanced jaw position, achieved by co-contracting the jaw opener and closer muscles without occlusal contact, might be an inherently safe reactive jaw position for reduction of head acceleration while the body experiences a predictable impact. Moreover, we found that jaw clenching does not contribute to the reduction of head acceleration in response to pendulum impact more than the distanced jaw position does. Most importantly, we found that the activation of DIG to minimise head acceleration in response to pendulum impact was similar in the clenching and no clenching conditions. This suggests that DIG may play a crucial role in the reduction of head acceleration, regardless of MA muscle activity.

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CONFLICTS OF INTEREST

The authors declare no conflicts of interest with respect to the authorship and/or publication of this article.

AUTHOR CONTRIBUTIONS

Yuto Tanaka and Yosuke Tomita designed the study; Kazuki Sako, Yuto Tanaka and Yosuke Tomita collected and analysed the data; Kazuki Sako and Yuto Tanaka wrote the paper; Yosuke Tomita, Tsuyoshi Yoshida, Yoshiaki Ono and Kosuke Kashiwagi revised the final manuscript. All authors have read and agreed to the published version of the manuscript.

PEER REVIEW

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DATA AVAILABILITY STATEMENT

The data that support the findings of this study are available on request from the corresponding author. The data are not publicly available due to privacy or ethical restrictions.

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REFERENCES

1. Hasegawa K, Takeda T, Nakajima K, et al. Does clenching reduce indirect head acceleration during rugby contact? *Dent Traumatol*. 2014;30:259-264.
2. Narimatsu K, Takeda T, Nakajima K, Konno M, Ozawa T, Ishigami K. Effect of clenching with a mouthguard on head acceleration during heading of a soccer ball. *Gen Dent*. 2015;63:41-46.
3. Collins CL, Fletcher EN, Fields SK, et al. Neck strength: a protective factor reducing risk for concussion in high school sports. *J Prim Prev*. 2014;35:309-319.
4. 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;42:566-576.
5. Caccese JB, Buckley TA, Tierney RT, et al. Head and neck size and neck strength predict linear and rotational acceleration during purposeful soccer heading. *Sports Biomech*. 2018;1:462-476.
6. Himejima A, Shirao K, Tsurumi A, Tanaka M, Morita S. Occlusal contact and muscle activity during judo. *Int J Sports Dent*. 2013;6:43-56.
7. Okazaki M, Tanaka M, Kawazoe T. Occlusal contact during offensive and defensive boxing. *Shika Igaku (J Osaka Odontol Soc)*. 2008;71:210-219. (in Japanese).
8. Sugihara D, Kawara M, Suzuki H, et al. Mandibular jaw movement and masticatory muscle activity during dynamic trunk exercise. *Dent J (Basel)*. 2020;8:132.
9. Lang NP, Bartold PM. Periodontal health. *J Clin Periodontol*. 2018;45:S9-S16.
10. Gonzalez YM, Schiffman E, Gordon SM, et al. Development of a brief and effective temporomandibular disorder pain screening questionnaire: reliability and validity. *J Am Dent Assoc*. 2011;142:1183-1191.
11. Harriss DJ, MacSween A, Atkinson G. Ethical standards in sport and exercise science research: 2020 update. *Int J Sports Med*. 2019;40:813-817.
12. Santos MJ, Aruin AS. Role of lateral muscles and body orientation in feedforward postural control. *Exp Brain Res*. 2008;184:547-559.
13. Santos MJ, Kanekar N, Aruin AS. The role of anticipatory postural adjustments in compensatory control of posture: 2. Biomechanical analysis. *J Electromyogr Kinesiol*. 2010;20:398-405.
14. Kanekar N, Aruin AS. Improvement of anticipatory postural adjustments for balance control: effect of a single training session. *J Electromyogr Kinesiol*. 2015;25:400-405.
15. Garrido garcia VC, Garcia cartagena A, Gonzalez sequeros O. Evaluation of occlusal contacts in maximum intercuspation using the T-Scan system. *J Oral Rehabil*. 1997;24:899-903.
16. Kerstein RB, Lowe M, Harty M, Radke J. A force reproduction analysis of two recording sensors of a computerized occlusal analysis system. *Cranio*. 2006;24:15-24.
17. Koos B, Godt A, Schille C, Göz G. Precision of an instrumentation-based method of analyzing occlusion and its resulting distribution of forces in the dental arch. *J Orofac Orthop*. 2010;71:403-410.
18. Chang YM, Shen YF, Hong HH, Liu CW. Using the T-scan III system to analyze occlusal function in mandibular reconstruction patients: a pilot study. *Biomed J*. 2015;38:52-57.
19. Throckmorton GS, Rasmussen J, Caloss R. Calibration of T-Scan sensors for recording bite forces in denture patients. *J Oral Rehabil*. 2009;36:636-643.
20. Cerna M, Ferreira R, Zaror C, Navarro P, Sandoval P. Validity and reliability of the T-Scan(®) III for measuring force under laboratory conditions. *J Oral Rehabil*. 2015;42:544-551.
21. Kanda Y. Investigation of the freely available easy-to-use software 'EZR' for medical statistics. *Bone Marrow Transplant*. 2013;48:452-458.
22. Faul F, Erdfelder E, Lang AG, Buchner A. G*Power 3: a flexible statistical power analysis program for the social, behavioral, and biomedical sciences. *Behav Res Methods*. 2007;39:175-191.
23. Faul F, Erdfelder E, Buchner A, Lang AG. Statistical power analyses using G*Power 3.1: tests for correlation and regression analyses. *Behav Res Methods*. 2009;41:1149-1160.
24. Fukumoto T, Tsurumi A, Tanaka M. Stomatognathic function during continuous physical activity in Nippon Kempo. *Int J Sports Dent*. 2016;9:58-71.
25. Miles TS, Flavel SC, Nordstrom MA. Control of human mandibular posture during locomotion. *J Physiol*. 2004;554:216-226.
26. Türker KS, Jenkins M. Reflex responses induced by tooth unloading. *J Neurophysiol*. 2000;84:1088-1092.

27. Schantz P, Randall-fox E, Hutchison W, Tydén A, Åstrand PO. Muscle fibre type distribution, muscle cross-sectional area and maximal voluntary strength in humans. *Acta Physiol Scand*. 1983;117:219-226.
28. Campbell KR, Warnica MJ, Levine IC, et al. Laboratory evaluation of the gForce Tracker™, a head impact kinematic measuring device for use in football helmets. *Ann Biomed Eng*. 2016;44:1246-1256.
29. Manoogian S, McNeely D, Duma S, Brolinson G, Greenwald R. Head acceleration is less than 10 percent of helmet acceleration in football impacts. *Biomed Sci Instrum*. 2006;42:383-388.
30. Jadischke R, Viano DC, Dau N, King AI, McCarthy J. On the accuracy of the head impact telemetry (HIT) system used in football helmets. *J Biomech*. 2013;46:2310-2315.
31. Winters J, DeMont R. Role of mouthguards in reducing mild traumatic brain injury/concussion incidence in high school football athletes. *Gen Dent*. 2014;62:34-38.
32. Ono Y, Tanaka Y, Sako K, Tanaka M, Fujimoto J. Association between sports-related concussion and mouthguard use among college sports players: a case-control study based on propensity score matching. *Int J Environ Res Public Health*. 2020;17:4493.
33. Takeda T, Ishigami K, Shintaro K, Nakajima K, Shimada A, Regner CW. The influence of impact object characteristics on impact force and force absorption by mouthguard material. *Dent Traumatol*. 2004;20:12-20.
34. Tanaka Y, Tsugawa T, Maeda Y. Effect of mouthguards on impact to the craniomandibular complex. *Dent Traumatol*. 2017;33:51-56.

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