Time Window of Head Impact Kinematics Measurement for Calculation of Brain Strain and Strain Rate in American Football

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Abstract—Wearable devices have been shown to effectively measure the head’s movement during impacts in sports like American football. When a head impact occurs, the device is triggered to collect and save the kinematic measurements during a predefined time window. Then, based on the collected kinematics, finite element (FE) head models can calculate brain strain and strain rate, which are used to evaluate the risk of mild traumatic brain injury. To find a time window that can provide a sufficient duration of kinematics for FE analysis, we investigated 118 on-field video-confirmed football head impacts collected by the Stanford Instrumented Mouthguard. The simulation results based on the kinematics truncated to a shorter time window were compared with the original to determine the minimum time window needed for football. Because the individual differences in brain geometry influence these calculations, we included six representative brain geometries and found that larger brains need a longer time window of kinematics for accurate calculation. Among the different sizes of brains, a pre-trigger time of 40 ms and a post-trigger time of 70 ms were found to yield calculations of brain strain and strain rate that were not significantly different from calculations using the original 200 ms time window recorded by the mouthguard. Therefore, approximately 110 ms is recommended for complete modeling of impacts for football.

Keywords—Instrumented mouthguard, Traumatic brain injury, American football, Time window, Head impact sensor.

INTRODUCTION

Mild traumatic brain injury (mTBI) is a global threat to human health.7 Sport-related repetitive head impacts, especially those received in American football, have the potential to cause mTBI3,33 and are associated with brain deficits41 and neurological changes.39 During a head impact, the skull rotates rapidly, and the brain’s inertia may cause brain deformation, which is associated with brain tissue pathology.15,39 Therefore, researchers have been investigating the relationship between head kinematics and brain deformation to predict the risk of injury.12,14,28,60,61

To measure head kinematics during on-field game play, several wearable devices have been developed. For sports with helmets, the Head Impact Telemetry System (HITS) has multiple accelerometers installed into the helmet and can provide 5 degrees-of-freedom (DoF) measurement of head movement.9,43,46,54 To have full head kinematics, an algorithm was developed to estimate 6 DoF head movement from 5 DoF measurement.35 In addition, individual sensor module that can be mounted to the outside of a helmet have also been developed.21 For sports without helmets, other solutions have emerged, including ear-mounted devices,24,42 skin-attached devices,22,38,44 and headband-mounted sensors.5,21 However, as the skin remains between the skull and skin-attached devices and headband-mounted sensors, the compliance of the skin...
has raised concerns that they cannot accurately replicate the movement of the skull. To overcome this limitation, researchers have employed instrumented mouthguards to measure head kinematics since the upper dentition is rigidly fixed to the skull. Recently, several instrumented mouthguards, containing accelerometers and gyroscopes, have been developed and used to collect on-field head impact data.\textsuperscript{2,6,8,13,18,21,30,45,49,53,58} To assess and improve the accuracy of the instrumented mouthguard\textsuperscript{2,6,26,30,49} the Hybrid III headform was modified to use as a surrogate in validation tests,\textsuperscript{46} and researchers are working on the challenges associated with instrumented mouthguards, e.g., the jaw slamming onto a mouthguard,\textsuperscript{27} the bandwidth of sensors,\textsuperscript{57} mechanical safety\textsuperscript{4} and false events (mouthguard is triggered when there is no real impact).\textsuperscript{8,13,25,36,56}

Despite these thorough investigations of head impact sensor performance, the minimum time window of measurements needed to accurately model brain kinematics is unknown. Once the mouthguard sensors are triggered, the head kinematics between the specific pre-trigger and post-trigger time will be recorded. As shown in Table 1, the pre-trigger and post-trigger times vary widely among different wearable devices. During a head impact, the sensor is typically triggered when the absolute value of linear acceleration at the accelerometer exceeds a threshold value. The 6-DoF kinematics between the sensor’s pre-trigger time and post-trigger time will subsequently be recorded. Then, several milliseconds are needed to save the data and reset the sensor for the next potential impact. To avoid missing important information from a head impact, the time window between pre-trigger and post-trigger should be long enough to include most of the variation of kinematic parameters to compute brain strain and strain rate peak. For some devices, this time window is adjustable by the user or can be easily modified by the manufacturer. In this study, we aim to provide the minimum time window of kinematic data needed to accurately model head impacts in American football.

Considering on-field impact loading is different from laboratory tests, we analyzed 118 on-field video-confirmed head impacts collected by Customized Stanford Instrumented Mouthguards (MiG2.0).\textsuperscript{6,8,30} The peaks of head kinematics (angular velocity, angular acceleration, and linear acceleration at the head center of gravity (CoG)) and the peak occurrence times were extracted. Then, 95th percentile maximal principal strain (95% MPS) peak and 95th percentile maximal principal strain rate (95% MPSR) peak were calculated by using Kungliga Tekniska Högskolan (KTH) finite element (FE) head model simulations.\textsuperscript{23} We applied a long time window of MiG2.0 measurement of head impacts as a reference (200 ms), and intentionally truncated time windows of MiG2.0 data to study the influence of the time window itself on the calculation of brain strain and strain rate. Two analyses were presented in this study according to the different purposes of using the mouthguard. To avoid the influence from the time window on the statistical distribution of calculated 95% MPS and 95% MPSR, we performed a pairwise comparison between the simulation results from original and truncated head kinematics. Furthermore, considering the need to capture the 95% MPS and 95% MPSR peaks in every event, we computed the history of head kinematics needed for accurate simulation and gave the range of the time needed for the calculations. Additionally, since 95% MPS has been found to vary according to different brains,\textsuperscript{29} we scaled the FE head model to six representative brain sizes\textsuperscript{55} to ensure that the time window was adequate for a large range of athletes.

### METHODS

**Collection of Head Impact Data**

The head impact data used in this study were collected by MiG2.0 devices (Fig. 1a) from the Stanford University football team’s training and games in 2019.\textsuperscript{30} The study was approved by Institutional Review Board at Stanford University (IRB). All impact events were confirmed by both manual video analysis and the neural network classifier.\textsuperscript{8} This analysis yielded a total of 118 head impact events and all events were subconcussive (no concussion was reported or diagnosed after any of the events). For each impact, the MiG2.0 recorded the angular velocities at 8000 Hz and the linear accelerations at 1000 Hz, both of which were filtered by 4th-order Butterworth low-pass filters with a cut-off frequency of 160 Hz.\textsuperscript{30} Then, the angular

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**TABLE 1. Time window of wearable devices to measure head kinematics.**

| Devices | Time Windows (ms) |
|---------|-------------------|
| Head Impact Telemetry System (HITS) | \([-8, 32]\textsuperscript{46,54} [-12, 28]\textsuperscript{59} |
| GForce Tracker (sensor module for helmet) | \([-8, 32]\textsuperscript{21} |
| SIM-G (sensor module for headband) | \([-10, 52]\textsuperscript{21} |
| X-patch | \([-10, 90]\textsuperscript{13} |
| Instrumented Mouthguard\textsuperscript{13} | \([-10, 50]\textsuperscript{13} |
| Instrumented Mouthpiece\textsuperscript{36} | \([-15, 45]\textsuperscript{15} |
| Vector Mouthguard | \([-16, 80]\textsuperscript{21} |
| Sports & Wellbeing Analysis Mouthguard | \([-1, 101]\textsuperscript{31} |
| Prevent Mouthguard | \([-10, 40]\textsuperscript{31} |
| Stanford Instrument Mouthguard (MiG) | \([-50, 150]\textsuperscript{31} |

\(t = 0\) ms is the device triggering.
acceleration was calculated by a 5-point derivative of the angular velocity and the linear acceleration was transformed from the accelerometer to head CoG according to the geometry measured from the Hybrid III Anthropomorphic Test Device 50th percentile Male headform. When the absolute value of linear acceleration at the sensor at any measuring direction is higher than 20 g, the mouthguard is triggered. 

\[ t = 0 \text{ ms is the MiG2.0 triggering, and } t < 0 \text{ and } t > 0 \text{ indicate the pre-trigger time and post-trigger time, respectively.} \]

**KTH Finite Element Head Model**

The KTH FE head model was developed by Royal Institute of Technology in Stockholm, Sweden (Fig. 1b). The model includes the scalp, the skull, the brain, the meninges, the cerebrospinal fluid and eleven pairs of the largest parasagittal bridging veins. This model was previously validated by cadaver head impact experiments. The skull was assumed to be rigid since stress waves will not pass through it and moved according to the measured angular velocity and linear acceleration. As a result, the brain would be deformed by the inertial force. Based on the simulation results, we first calculated the 95% MPS and 95% MPSR at each time point, and then determined the peak values over time. For all simulations presented in this study, the double precision solver was used.

**Individual Head Models**

To generalize these results to a wide range of athletes, six representative brains in the WU-Minn Human Connectome Project (WUM HCP) were used to study the effect of brain geometry. Since brain strain has been shown to be more dependent on rotation instead of translation, we scaled the node coordinates of the original KTH model according to the moment of inertia instead of the brain’s mass. Three male brains (largest, 50th percentile, 5th percentile) and three female brains (50th percentile, 5th percentile, smallest) were used in this study. For each representative brain, the original KTH model was scaled to have the same moment of inertia about the X axis (posterior to anterior), Y axis (left to right) and Z axis (superior to inferior) with the origin at the brain’s CoG. The scaling ratios are listed in Table 2. All analyses were performed on the seven FE head models (KTH original + six representative head models).

**Time Window that Yields Non-significantly Change of 95% MPS and 95% MPSR Peaks**

As shown in Table 1, head impact sensors have different pre-trigger and post-trigger times. To investigate how these differences influence brain strain analyses, we performed simulations with intentionally truncated time windows of head kinematics and compared the results with the original brain strain (calculated by the original 200 ms time window of head kinematics). For the pre-trigger time, we started the simulation at −30 ms, −20 ms, −10 ms, and 0 ms (triggering) and calculated the peaks of 95% MPS and 95% MPSR at 30 ms, 40 ms, 50 ms, 60 ms, and 70 ms. The pre-trigger time and post-trigger time were varied together, and the simulation results were compared with that calculated by the original head kinematics (−50 ms, 150 ms). Each group of 95% MPS peaks and 95% MPSR peaks with the same pre-trigger or post-trigger time was tested by the Anderson-Darling test for normality, and then compared with original brain strain by pairwise t-test.

**History and Time Range Needed to Accurately Calculate 95% MPS and 95% MPSR Peaks**

In addition to the statistical significance of difference for all impacts as a group, we also investigated the

![FIGURE 1. (a) Stanford Instrumented Mouthguard (MiG2.0). (b) KTH finite element head model.](image)
influence of time window for each impact separately. Due to the viscoelasticity of the brain tissue, the 95% MPS and 95% MPSR peaks rely on the history of head kinematics before the peak time. To investigate the history needed to calculate the accurate peaks, we truncated the time window of the original kinematics individually. For each impact, a series of simulation cases were performed with the kinematics truncated from the original. For the impacts in which the peak time was after −10 ms (more than 40 ms before the peak was captured, since the original pre-trigger time was −50 ms), we truncated at every 2 ms for a total of 40 ms before its peak times. For the impacts in which the peak time was before -10 ms (less than 40 ms before the peak was captured), we truncated kinematics at every 2 ms until the beginning of the data. An example of the calculation of the history needed was shown in Fig. 2. In this event, the peak time for 95% MPS was 5 ms, and we truncated the kinematics at 3 ms, 1 ms… until −35 ms and performed simulations with every truncated kinematics. Cases A–D (Figs. 2a–2d) are the 95% MPS with the kinematics truncated at 10 ms, 12 ms, 14 ms, 16 ms. Then, the relative error in each truncated kinematics case ($e_i$) was calculated as,

$$e_i = (X_i - X_0)/X_0$$

where $X_i$ is the 95% MPS peak in the truncated kinematics case $i$, and $X_0$ is 95% MPS peak calculated by the original kinematics. We adopted the critical relative error of 6.8% ($e_{crit}$) to define the accurate 95% MPS peak measurement according to previous studies about the propagation of the kinematics measurement error to brain strain. Then, the history needed for 95% MPS peak was calculated by the interpolation as,

$$\text{History} = \begin{cases} 
(X_B - (1 + e_{crit}) \cdot X_0)/(X_B - X_C) \cdot 2\text{ms for } X_B > X_0 \\
(X_B - (1 - e_{crit}) \cdot X_0)/(X_B - X_C) \cdot 2\text{ms for } X_B < X_0 
\end{cases}$$

where the truncated kinematics case B included the longest history among the inaccurate cases (Fig. 2b), and truncated kinematics case C was the simulations starting 2 ms after B (Fig. 2c). For the events in which 40 ms was not enough, we extended the history to include the range from 40 ms to 60 ms. The same process was performed for 95% MPSR and $e_{crit}$ of 6.8% was adopted to define the accurate measurements. Then, the time range needed for the accurate calculation is defined with the peak time of the 95% MPS or 95% MPSR as,

$$\text{Time Range} = [\text{Peak Time} - \text{History, Peak Time}]$$

### RESULTS

#### Peaks of Head Kinematics, 95% MPS and 95% MPSR

For the 118 video-confirmed football head impact events, the peak values of angular acceleration (Fig. 3a), angular velocity (Fig. 3b) and linear acceleration at CoG (Fig. 3c) are plotted against their corresponding peak times. Kernel density estimations are plotted at the top. The 95th percentile of peak values (95% peak), the 5th and 95th percentile of peak time (5% and 95% peak time, respectively) are also plotted. In Figs. 3a and 3b, the peak time for angular acceleration and linear acceleration at CoG are similar and close to the triggering, while that of the angular velocity is more dispersive between $t = -10$ ms and $t = 50$ ms. It should also be noted that the kinematics of a few cases reached peaks much before or after the triggering.

To calculate the 95% MPS and 95% MPSR, the brain strain was calculated by the FE models of seven brain geometries with the kinematics of the 118 head impact events. Three example impact events are plotted in Fig. 4. The raw linear acceleration (trigger), the processed linear acceleration at CoG and the processed angular acceleration are given in Figs. 4a–4c. Although the waveforms of traces are similar among different head models, the value of 95% MPS and 95% MPSR varied considerably. Multiple local maximum points can be observed in 95% MPS and 95% MPSR traces, and the peak one varies among different head models in some cases. For example, in impact event 1 (Figs. 4e1 and 4e1), 95% MPS and 95% MPSR reach the peak similarly among different head models. However, 95% MPSR in impact event 2 (Figs. 4d2 and 4e2) and 95% MPS in impact event 3 (Figs. 4d3 and 4e3) reach peaks at different times. The peak values and the peak times are plotted in Fig. 5. Both the 5th and the 95th percentile peak time of 95% MPS are earlier than that of 95% MPSR, and a small difference in peak time among brains can be observed. Similar to
the kinematics, the 95% MPS and 95% MPSR peaks far before or after the triggering in a few cases.

**Influence of Pre-Trigger and Post-Trigger Time in 118 Football Head Impact Events**

The head kinematics histories were truncated to $-40$ ms, $-30$ ms, $-20$ ms, $-10$ ms, and 0 ms for the pre-trigger time and to 30 ms, 40 ms, 50 ms, 60 ms and 70 ms for the post-trigger time. The pre-trigger and post-trigger time were varied together, and the simulation of the results were compared with the original time window ($[-50$ ms, 150 ms]). Each group truncated by the same time windows was confirmed to follow a normal distribution by the Anderson-Darling test and was then compared using pairwise t-test with the simulation results by the original time window of head kinematics. The p-value of the pairwise test was shown in Fig. 6, in which the heatmap shows if the pre-trigger and post-trigger time yield significantly change of simulation results from the original for each head. The comparison showed that the pre-trigger time of $-40$ ms and post-trigger time of 70 ms are able to give non-significantly different 95% MPS peaks (Fig. 6a) and 95% MPSR peaks (Fig. 6b) for all head models.

**History Needed and Time Range for Accurate Analysis**

Although the time window for statistical non-significant difference was found to be $[-40$ ms, 70 ms], we also investigated the history and time ranges needed for each event, and the time range needed in some events might be outside the time window of $[-40$ ms, 70 ms]. Among different head models, the 95th percentile of history needed varied from 23 to 37 ms for 95% MPS peak and from 20 to 31 ms for 95% MPSR, and longer history was needed for larger brains (Fig. 7, Table 2). Then, for each event, the history and the

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**FIGURE 2.** Example calculation of the history needed for 95% MPS peak. (a)–(d) Trace of 95% MPS calculated by the kinematics truncated at 10 ms, 12 ms, 14 ms, 16 ms before the peak time, respectively. (e) 95% MPS peak calculated by the kinematics shortened at every 2 ms to 40 ms before the peak time. (a)–(d) is indicated in (e), and the history needed is calculated by the interpolation between cases B and C.

**FIGURE 3.** The head kinematics peaks and the time for the peaks of 118 on-field football impacts. The vertical dashed line is the trigger time ($t = 0$ ms). The left and right vertical dotted lines are the 5th percentile and the 95th percentile of the time for the peaks, respectively, and the horizontal dotted line is the 95th percentile of the head kinematics peaks. In a few cases, the peaks happen far away from the trigger.
peak time were taken into Eq. 3, and the time range needed for accurate calculation is given in Fig. 8. The time ranges for most head models were within $[240 \text{ ms}, 70 \text{ ms}]$. In a few events with late peak time, because of the accumulation of numerical error in a long-time simulation with the original data, the relative error of 95% MPS peak was larger than $\epsilon_{\text{crit}} = 6.8\%$ when 60 ms history was included in simulations, so these cases were not plotted in Fig. 7a and their peak time were plotted as gray bars in Fig. 8a.

FIGURE 4. Three example head impacts. (a) raw linear acceleration at sensor which triggers the mouthguard (the triggering threshold is 20 g, X: back to front, Y: left to right, Z: top to bottom); (b) processed linear acceleration at center of gravity (CoG) of head; (c) processed angular acceleration of head; (d) 95% MPS calculated by seven heads models; (e) 95% MPSR calculated by seven heads models. In the impact 1, 95% MPS and 95% MPSR reach the peak closely among different head models. The 95% MPSR in the impact 2 and the 95% MPS in impact 3 have different peak time among heads.
Time Window of Head Impact Kinematics Measurement
FIGURE 5. (a) The 95% MPS peaks and the time for the peaks; (b) the 95% MPSR peaks and the time for the peaks. (1–7) KTH original and six representative brains. The vertical dashed line was the trigger time \( t = 0 \) ms. The left and right vertical dotted lines was the 5th percentile and the 95th percentile of the time for the peaks, respectively, and the horizontal dotted line is the 95th percentile of the 95% MPS or MPSR peaks. The Peak time and value varied among the brains. The peak time of most events was close to \( t = 0 \) ms, but the peak time of a few events were far before or after \( t = 0 \) ms.

DISCUSSION

In this study, we investigated how the time window of instrumented mouthguard measurements influences the further analysis of 95% MPS and 95% MPSR in 118 American football head impact events. In contrast to laboratory testing, where there is only one local maximum point near the device triggering, 30 on-field data can have multiple local maximum points (Fig. 4). Most peaks of kinematics and brain strain occur when the device is triggered, and a few events had peaks much earlier or later. It was also found that peak time for 95% MPS and 95% MPSR are influenced by brain geometry (Fig. 5). To address this, seven head models, including the original KTH head model and six representative head models, were used in simulations to observe the effect of brain size on the kinematics. Larger brains were found to require a longer history of kinematics for accurate calculation (Fig. 7). This result is likely due to the increase in inertia and the increased distances stress waves are required to travel in larger brains during similar impact conditions. When considering all head models, a time window of \([-40 \) ms, 70 ms\] was found to give 95% MPS and 95% MPSR peaks that were not significantly changed from the original data (Fig. 6). Therefore, \([-40 \) ms, 70 ms\] is suggested as the minimum time window for instrumented mouthguards. Furthermore, we computed the time range that is needed to calculate the accurate 95% MPS and 95% MPSR peaks for every event. We found that the kinematics outside this range \([-40 \) ms, 70 ms\] were needed in some events (Fig. 8), so a wider time window will help to capture the peaks in every event. According to Table 1, most of the devices have time window shorter than 90 ms except X-patch, Sports & Wellbeing Analysis mouthguard and Stanford Instrumented Mouthguard.

The instrumented mouthguard in the present study is triggered by linear acceleration collected by an accelerometer, and subsequently records head kinematics in a specified time window as an impact event. It should be noted that linear acceleration varies with location for a rigid body. In the MiG2.0 mouthguard, the accelerometer is located at the incisor (Fig. 1a) in consideration of reducing the influence of jaw slamming. 27 Therefore, in this study, \( t = 0 \) ms represents when the linear acceleration at the incisor reaches the triggering condition. In other mouthguards with an accelerometer at the molar, 13,30 and other types of devices, 5,9,22,38,42–44,46,54 the angular velocity and angular acceleration may shift because \( t = 0 \) ms corresponds to different triggering conditions. Therefore, the adequate pre-trigger time and post-trigger time may shift in other devices, but the length of the time window should be the same.

Accelerometers are used to trigger instrumented mouthguard sensors instead of gyroscopes as they measure angular velocity, and high angular velocity of the head may occur without an impact. 17 As a result, high-level angular acceleration may occur much earlier or later than the triggering. Since head rotation deforms the brain, 19,20 95% MPS peak and 95% MPSR peak may also occur much earlier or later than the triggering. Early peak times can be found in kinematic parameters, 95% MPS, and 95% MPSR. Peaks of linear acceleration at CoG were also found before the triggering because of the transformation from accelerometer (at the incisor) to the head CoG. Furthermore, because brain tissue is history-dependent, 11,23 head kinematics before the peak time are needed to calculate an accurate peak value. Therefore, the pre-trigger time should be long enough for the analysis of 95% MPS and 95% MPSR. Late peak time can also be found in kinematic parameters, 95% MPS, and 95% MPSR. An example with late peak time is shown in Fig. 4a4, where we find two local maximum points of angular acceleration far from each other. The linear acceleration associated with the first angular acceleration local maximum point triggered the mouthguard and the second local maximum point is the highest, which made 95% MPS and 95% MPSR reach peaks at the second local maximum point in most head models (Figs. 4b4 and 4c4). Multiple instances of head contact within a short time may be a potential explanation for this finding. For example, it is possible that two head impacts caused two peaks in Fig. 4c3, respectively, and the kinematics resulting from each contact overlapped. Another potential
Time Window of Head Impact Kinematics Measurement

Original

Largest Male

50th Perc. Male

5th Perc. Male

50th Perc. Female

5th Perc. Female

Smallest Female

(a1) MPS

(b1) MPSR

(a2)

(b2)

(a3)

(b3)

(a4)

(b4)

(a5)

(b5)

(a6)

(b6)

(a7)

(b7)
explanation of this is rebounding in one contact. The kinematics traces alone are not sufficient to find out the reason for multiple local maximum points because there is not a period that all kinematics remain steady zero. A clear video of each impact process is needed. However, in this study, limited by the frame rate, resolution, and camera angle of videos, we could not investigate the reason for these multiple local maximum points of the kinematic traces, and we believe that lab-reconstruction of head impacts are needed to provide more insight into this in the future. As a result, the entire variation of kinematics should be included to calculate 95% MPS and 95% MPSR. A potential alternative of long post-trigger time is to have two subsequent recorded events and merge the head kinematics for the analysis. However, in the current device design, the sensors and microcontroller need several milliseconds to save data and reset for the next event, and the kinematics during this time will be missed. As shown in Fig. 8, it is possible that the missing kinematics are needed to accurately calculate brain strain, so this approach may either miss the peak or cause inaccuracy.

When the kinematics are truncated by pre-trigger and post-trigger time, the pre-trigger time will influence the results by changing the history of kinematics before the peak time, and the truncated post-trigger time will influence the 95% MPS and MPSR peak by missing the peak time. The effects of truncated pre-trigger time and post-trigger time will be combined when compared with the original simulation results. By comparing the results of different pre-trigger and post-trigger time, [−40 ms, 70 ms] was found to not significantly change 95% MPS and MPSR peaks for all seven brain geometries. However, the head kinematics outside [−40 ms, 70 ms] were needed to calculate the accurate value for brain strain in a few impact events (Fig. 8). As we expect more data will be collected by MiG2.0 mouthguards in the future, the number of peaks that happen outside [−40 ms, 70 ms] should be examined.

In the present study, we observed multiple local maximum points within a single head impact event, and found that the kinematics peak, 95% MPS peak, and 95% MPSR peak may correspond to different local maximum points. For example, as shown in

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**FIGURE 7.** Histogram of the history needed for accurate calculation (within critical relative error of 6.8%) of 95% MPS peak (a) and 95% MPSR peak (b). (1–7) KTH original and six representative brains. In each plot, the curves are the kernel density, and the left and right vertical dotted lines are the mean and the 95th percentile history needed. From a2–a7 and b2–b7, the history needed increases when the size of head decreases (Perc.: Percentile).
Figs. 4a3, 4b3, and 4c3, the angular acceleration reached its peak at the second local maximum point, but the 95% MPS for the largest male head model reached its peak at the first local maximum point. This phenomenon shows that the whole variation of angular velocity should be recorded and raises concerns about some brain injury metrics\textsuperscript{23,47,51,52,59} and machine-learning head models\textsuperscript{62} that predict brain strain based on the peak values of kinematic parameters. When these metrics are applied to head impacts with multiple local maximum points, the kinematics at one local maximum point may be used to predict brain strain at another, thus decreasing the prediction accuracy. Furthermore, multiple local maximum points will also increase the duration that brain strain sustains at a high level. When comparing impact events 1 and 2 in Fig. 4, although the 95% MPS peaks were close, three local maximum points can be found in event 2 while only one can be found in event 1. As a result, 95% MPS in event 2 sustains at a high level longer than in event 1. Although we use the peak values of brain strain and strain rate to evaluate the risk of injury, we do not know if injury accumulates with long duration. Recently, repetitive head impacts have been suggested to cause more neurological sequelae\textsuperscript{1,10,32,34,37}, which indicates that injury accumulates with multiple impacts. Therefore, the effect of multiple local maximum points and the duration of high brain strain and strain rate should be further investigated.

Brain geometry has been found to influence brain displacement\textsuperscript{63}, and thus brain strain as well. Recently, with the development of subject-specific head models\textsuperscript{29}, it is clear that brain geometry will influence both the peak value and the peak time of brain strain. The large difference of the 95% MPS and the 95% MPSR indicate that population-specific or customized FE head models are needed to accurately reconstruct the mechanical response of the brain. As previously mentioned, multiple local maximum points of kinematics
exist in one on-field head impact event, and the one causing the largest 95% MPS and 95% MPSR varies according to brain geometry. The history needed for accurate calculation also varies with brain geometry. Based on Table 2 and Fig. 7, larger head models need longer histories of kinematics for calculation of both 95% MPS peak and 95% MPSR peak. Therefore, the peak time varies among different head models. Considering that the time windows of mouthguard measurements should be suitable for all athletes, we chose the time window of $[-40 \text{ ms}, 70 \text{ ms}]$, which will still avoid causing significant differences from the original time window of data for all head models. Furthermore, the different responses of 95% MPS and 95% MPSR also indicate that a subject-specific brain injury metric is warranted. The 95% MPS peak given by FE head model was used to decide the parameters of brain injury metrics.\textsuperscript{12,14,28,50} However, as shown in Figs. 3 and 4, the 95% MPS peak in the largest male head model is much larger than that in the smallest female. As a result, the current injury threshold may miss some actual injury cases.

This study has several limitations. First, although different head models were adopted, the individual differences in the brain tissue properties were not considered. The material model and parameters will influence the history needed for accurate calculation, and then influence the time window. In the KTH model,\textsuperscript{23} the Ogden hyperelasticity model\textsuperscript{40} was assumed to model the brain tissue, and the parameters were obtained from these experiments.\textsuperscript{11} With more knowledge of the individual differences in brain tissue, future research should be performed to include the influence of material properties on necessary time windows. Regarding brain geometry, scaling the model according to moment of inertia may not accurately represent the influence of individual differences. We expect the development of the method of mesh morphology or remeshing will lead to adopting individual difference in the future. Second, the history needed for an accurate 95% MPS peak was not calculated in a few cases because the relative error was higher than 6.8% although 60 ms history was included in the simulation. This might be caused by the accumulation of numerical error after long-time simulations and need further investigation. Lastly, the 118 head impact events used in this study were collected by the MiG2.0 mouthguard, which has a time window of $[-50 \text{ ms}, 150 \text{ ms}]$. To the best of our knowledge, this is the longest time window of any head impact sensor, and we found that data truncated to a time window of $[-40 \text{ ms}, 70 \text{ ms}]$ will give results that are not significantly different from the original time window. However, it is possible that this dataset does not reflect the actual distribution of peak times in American football, as peaks may still occur before or after the $[-50 \text{ ms}, 150 \text{ ms}]$ time window. Furthermore, the current time window was decided based on the MPS and MPSR peaks metrics. In the future, the minimum time window may be modified if we find other mechanical metrics that are more clinically relevant to TBI. Moreover, the current study was based on American football head impact data. For other sports like soccer, martial arts, or ice hockey, for example, further studies are needed to find the appropriate minimum time windows because of different impact types and different head protection equipment.

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