Computational Prediction of Mandibular Muscle Forces During Sporting Head Impacts

Eleni Koumantou, John E. Lloyd, Benedikt Sagl, Mazdak Ghajari

I. INTRODUCTION

Traumatic brain injury (TBI) in contact sports has become a major public health concern, with authorities calling for urgent action to monitor and reduce head impact exposure in athletes. Instrumented mouthguards (iMGs) have now become an indispensable part of the monitoring system [1].

An iMG is a wearable sensing device with integrated gyroscopes and accelerometers, coupled to the skull through the upper dentition, allowing it to measure head kinematic traces during a head acceleration event (HAE). Validation studies of iMGs' measurements have produced contradictory results [2–5], partially because they use different experimental methodologies, particularly in how they simulate the interaction between the mandible and upper dentition, which has large effects on iMG's measurements. Currently, the response of the mandible during HAEs and its interaction with the iMG are largely unknown.

Understanding this interaction can be informed by a computational model that predicts mandibular response under realistic impact conditions. Musculoskeletal (MSK) jaw models have been developed for studying chewing and wide jaw opening [6–7], but they have not been used in high acceleration events. Here we apply these models to sporting HAEs to predict mandible response and to provide an estimation of the dynamic loading generated by muscles connected to the mandible.

II. METHODS

We used the open access jaw model of Hannam *et al.* [6][8], developed using the open-source ArtiSynth Modeling Toolkit [9]. It includes rigid bodies for the cranium, maxilla, mandible, and hyoid, represented with CT-derived surface meshes. Twelve bilateral muscle groups are included, described as Peck-type muscles [7], a model similar to Hill-type muscles but with linear passive force-length properties, empirically verified for jaw muscles. The temporomandibular joint (TMJ) is represented by bilateral planar constraints, angled 40° downward and 15° forward, restricting the translation of the mandibular to the planar surface, which approximates the curvilinear condylar path [6]. A planar surface perpendicular to this prevents any posterior movement of the anatomical centre of the condyle from its initial position, while a third planar constraint prevents lateral motion. To simulate head acceleration events we extended the model in two ways: (i) a custom SkullMotionController reads time-stamped position and orientation data at 1 kHz from the iMG-derived trajectory and applies them as relative transformations to the cranium and maxilla; and (ii) the bilateral TMJ planar constraints are re-registered with the skull controller so they track the cranium throughout the impact. Direct contact loading on the head or face from the impactor is not modelled; the iMG-derived kinematics are applied only to the cranium and maxilla, and the mandible responds through inertial coupling, the TMJ capsule, and (where active) the muscles. For this preliminary study, a passive scenario with zero muscle excitation was simulated, similar to cadaver experiments [10].

The ConstrainedBackwardEuler scheme was used for integrations at 1 ms timesteps. We simulated representative impact trajectories sampled from Mixed Martial Arts (MMA), boxing, and Rugby Football Union (RFU) iMG datasets, covering a range of impact types. Representative traces were drawn from an unsupervised clustering pipeline that applies shape-based distance on linear-acceleration and angular-velocity channels, with the number of medoids per dataset selected by an unsupervised quality score.

III. INITIAL FINDINGS

During a 50 ms settling phase, the jaw reached a stable equilibrium supported entirely by the TMJ capsule and articular constraints in all cases. Inspection of simulation videos confirmed a physically plausible mandibular inertial response. Bilateral asymmetry was observed in unilateral impacts, consistent with the expected directional response for an asymmetric input. Peak muscle forces varied across impact and muscle types (Fig. 1). Jaw-closing muscles (temporalis, masseter, medial pterygoid) consistently produced higher mean forces than jaw-opening muscles (lateral pterygoid, anterior digastric, mylohyoid, geniohyoid): 0.72 ± 0.14 N vs 0.11 ± 0.06 N.

E. K. Author is a PhD student in Biomechanics at Imperial College London in London, UK (e.koumantou24@imperial.ac.uk , +44 7842059979). J. L. Author is a Research Associate of Computational Biomechanics in the Department of Electrical and Computer Engineering at the University of British Columbia in Vancouver, CA. B. S. Author is the Head of Competence Center Artificial Intelligence in Dentistry at the Medical University of Vienna, AT. M. G. is an Associate Prof of Brain Biomechanics at Imperial College London.

Jaw-closing muscles reached peak force within 18–43 ms of impact onset, approximately 20 ms before the jaw-opening muscles, indicating a sequential loading pattern.

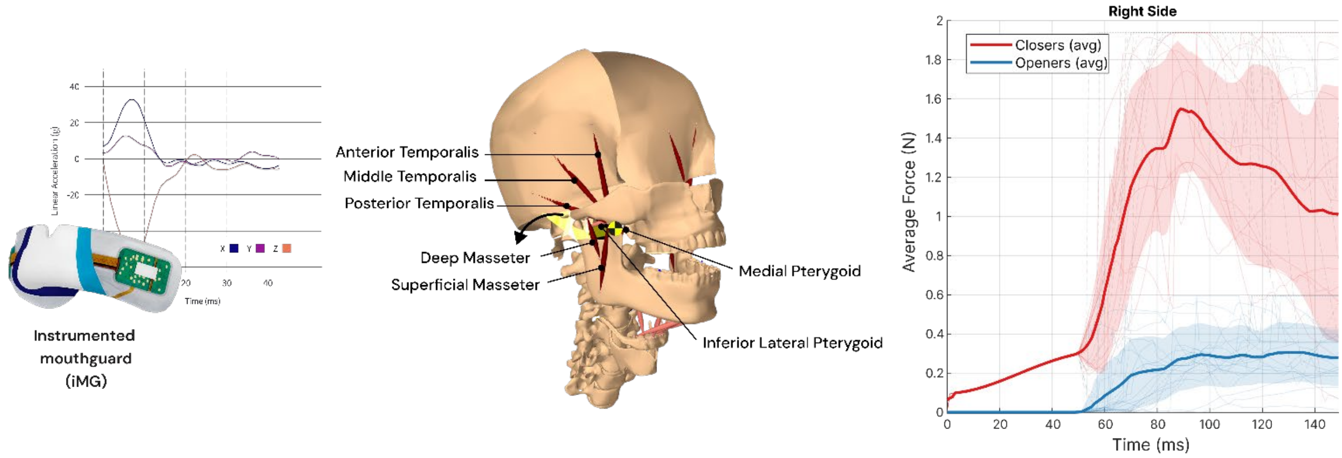

Fig. 1. From left to right: Example of kinematic traces from an iMG that guide the head motion; An oblique view of the model at a later timestamp during the impact — the control surfaces are shown as semi-transparent rectangles and the muscles have been annotated; A comparative plot of the normalised forces generated in the closers vs openers during passive impact scenarios.

IV. DISCUSSION

We used multibody dynamics to predict mandible motion and muscle forces during head kinematics in contact sports. While a similar approach has been applied to other body regions under dynamic loading, particularly the neck [11], no prior study has investigated mandible and its muscles reactions in sporting HAEs. We applied head kinematics measured with iMGs to a detailed and validated musculoskeletal model with 24 muscles, allowing us to provide first insights into mandibular motion and individual muscle forces. We show that closing muscles sustain larger forces than opening muscles, indicating their importance for developing biofidelic headforms that can be used for the evaluation of iMGs.

Jaw models are inherently difficult to validate because of the anatomical restrictions of the region [6,12]. We observed plausible mandible motion, with peak passive forces of 0.99 ± 0.22 N for the jaw closers — consistent with the Peck-type material's passive-stretch contribution (1.5% of F_{max} per muscle [7]) and approximately two orders of magnitude below the maximum isometric force of any individual closer. This indicates that even in a fully cadaveric scenario, passive masticatory tissue transmits a measurable inertial load to the mandible-cranium coupling, which is the lower-bound noise floor for an iMG measurement.

A limitation of this study is that iMG-derived kinematics were applied as prescribed cranial motion, with the mandible responding purely through inertial coupling; impacts that strike the mandible directly are not modelled as such. We simulated passive muscle, representing the cadaveric response. Future work will use the model to study: i) maximum voluntary clenching, where all closer muscles will be activated, simulating bracing for an incoming impact, and ii) minimum muscle activation patterns required to maintain incisor contact throughout the impact duration using an inverse simulation. Additionally, we will use representative traces from a wider range of impact data to quantify the protective role of voluntary pre-activation and the minimum reflexive response needed to maintain a clenched mandibular state. This preliminary study provides first insight into mandible and muscle response to head kinematics in sports, which can inform future development of more biofidelic methods for testing iMGs.

V. REFERENCES

[1] Jones B *et al.,* Br J Sports Med, 2022. [2] Siegmund GP *et al.,* Ann Biomed Eng, 2015. [3] Liu Y *et al.,* Ann Biomed Eng, 2020. [4] Gabler L *et al.,* Ann Biomed Eng, 2022. [5] Abrams MZ *et al.,* Ann Biomed Eng, 2024. [6] Hannam A *et al.,* J Biomech, 2008. [7] Peck C *et al.,* Arch Oral Biol, 2000. [8] Stavness I *et al.,* Biomed Simul, 2006. [9] Lloyd JE *et al.,* Soft Tissue Biomech Model Comput Assist Surg, 2012. [10] Venkatraman J *et al.,* Ann Biomed Eng, 2024. [11] Holzinger T *et al.,* Comput Methods Programs Biomed, 2024. [12] de Zee M *et al.,* J Biomech, 2007.