Preliminary Facial Impact Model Evaluation for Micromobility Injury Assessment

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I. INTRODUCTION

Micromobility vehicles such as e-scooters continue to gain a larger share of urban transportation systems, facilitated by the widespread availability of shared services. This growth has been accompanied by a rise in associated injuries, placing an increasing burden on the healthcare systems [1]. As vulnerable road users (VRUs), micromobility riders frequently sustain injuries to the head, face, and extremities, often requiring prolonged recovery and sometimes resulting in long-term functional impairments [2–3]. Understanding the underlying injury mechanisms is essential for developing preventive measures in vehicle design, infrastructure, and policymaking. Finite element human body models (HBMs) offer strong potential to address this challenge; however, the biofidelity of key regions remains unvalidated and may require improvement before being applied to micromobility crash scenarios.

Facial fractures are common among micromobility riders, particularly e-scooter users [4–5]. Nasal fractures are the most frequent, followed by mandibular and zygomaticomaxillary fractures [6]. These patterns highlight the importance of capturing localised tissue response, motivating injury prediction using a tissue-based risk function. As an initial step, this study evaluates the biofidelity of the face and head components of the VIVA+ Human Body Models [7] under a facial impact condition targeting the nose.

II. METHODS

Submodels of the face and head of the VIVA+ (v2.0.1) 50th percentile female (50F) and male (50M) models served as the basis for this work. Young's moduli of cortical and trabecular bone in the maxilla, zygoma, and mandible were set to 9.1 GPa [8] and 53 MPa [9], respectively. Cortical thickness in the maxilla and zygoma was set to 1.5 mm in most regions [10], decreasing to 0.5 mm at the nasomaxillary buttress [10] and 1.2 mm at the nasal cavity [11]. Nasal cartilage was represented using the LS-DYNA MAT077 Ogden-Rubber (first-order) material model with μ_1 = 0.01274 MPa and α_1 = 28.8 [12]. The developed face submodels and facial impact boundary conditions are available at https://openvt.eu/vivaplus/publications/abe-2026-face-impact.

The submodels were subjected to facial impact conditions (Fig. 1) derived from Nyquist *et al*. (1986) [13–14] and simulated using LS-DYNA (12.2.2 SMP, Ansys Inc., USA). The Frankfort plane was oriented horizontally, with the impactor height was aligned with the infraorbital margins. The unsupported, free-floating submodels were impacted by a rigid cylindrical impactor directed horizontally toward the nasal and zygomaticomaxillary regions. Two impactor masses were used (32 and 64 kg), with corresponding average contact velocities of 5.3 and 2.9 m/s, respectively [13]. Contact force and horizontal displacement at the impactor midpoint were extracted and compared with experimental force–displacement data. Contact force data were filtered using CFC600.

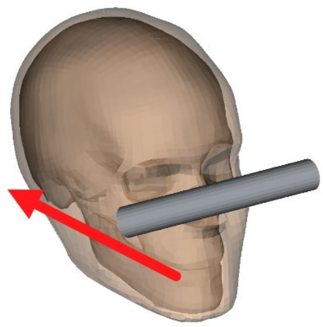

Fig.1. VIVA+ 50F head and face submodel under facial impact

III. INITIAL FINDINGS

For the 32 kg impactor (Fig. 2), the VIVA+ 50F and 50M models were stiffer than the experimental curves [13], with slopes of 4.3 and 4.2 kN/cm, respectively (0.25–3.5 kN). In contrast, the 64 kg impactor responses showed

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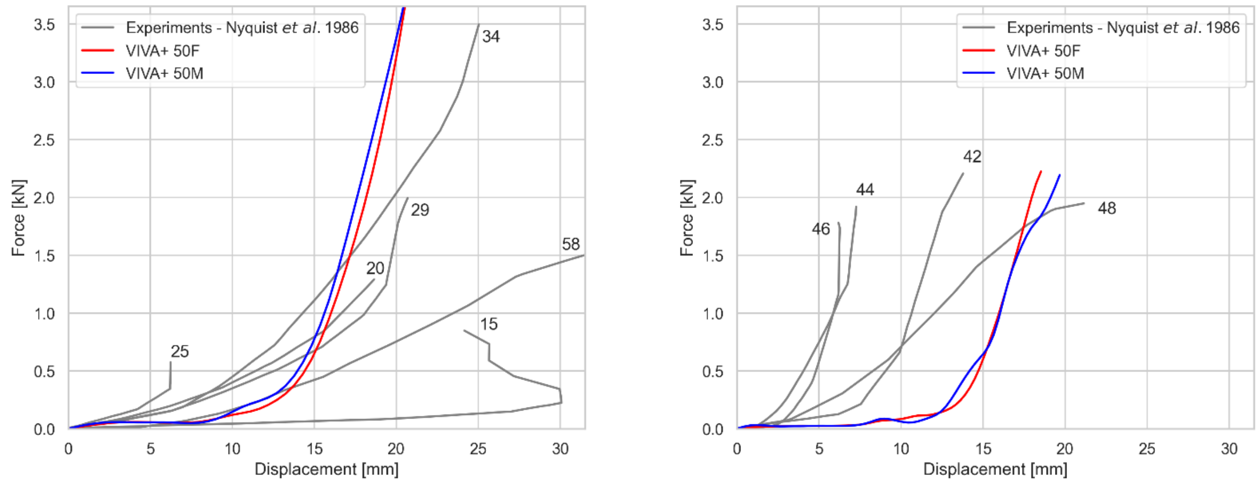
good agreement with the experimental data, with slopes of 4.4 and 3.1 kN/cm, respectively (0.25–2.2 kN).



(a) 32 kg impactor, contact velocity 5.3 m/s (b) 64kg impactor, contact velocity 2.9 m/s

Fig. 2. Comparison of force-displacement curves from VIVA+ simulations and the experiments by Nyquist *et al.* (1986) [13]. Numbers at the end of the grey curves indicate subject IDs from the original study.

IV. DISCUSSION

The force-displacement responses of the VIVA+ simulations (50F and 50M) exhibited a concave-up curve, with a steeper slope above ~0.25 kN. The initial low-stiffness region (<0.25 kN) is likely due to deformation of the nasal soft tissues and bone. The variation in this region is attributed to inter-individual differences in facial morphology (e.g., nasal height, cartilage structure, soft tissue thickness), consistent with previous studies [13–15].

Simulation responses for the 32 kg impactor were stiffer than the experimental data [13]. Strain rate dependent viscoplastic material properties derived from cranial cortical bone [16] were applied to both facial and cranial cortical bone. At higher impact velocities, strain-rate dependent material behavior of soft tissues and cortical bone, together with increased inertial effects, likely contribute to a steeper force–displacement response. These factors warrant further investigation to improve VIVA+ HBM head and face biofidelity. The model captured the overall force-displacement response but overestimated facial stiffness. As the model is intended for facial injury prediction, the observed stiffer response may affect prediction accuracy. Improving facial bone material properties is therefore important to enhance model biofidelity and injury prediction accuracy.

A limitation of this study is that only a single facial impact condition was investigated. Future work would consider impacts to other regions, such as the frontal and maxillary areas, as examined by Allsop *et al*. (1988) [15]. In addition, lateral chin impact conditions, representing chin–ground contact [17], should be studied, as these are likely relevant to micromobility crashes. More broadly, future work will continue validating and implementing tissue-based injury prediction methods for other body regions of the VIVA+ HBM (e.g., extremities), towards applying the model to micromobility crashes.

V. ACKNOWLEDGEMENTS

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