The Influence of Neck Posture on Cervical Spine Response During Head-First Impact Using a Human Body Model

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I. INTRODUCTION

Head-first impact (HFI) loading has been reported to cause cervical spine (CS) injuries, which often incur significant personal, financial and societal costs [1]. Specifically, the CS experiences bony and ligamentous injuries as a result of the compressive force upon impact and the compressive momentum of the following torso. HFI loading is found in several scenarios, including vehicle rollovers, falls and sports [2]. There are several factors reported to affect the response of the CS during HFI, such as impact velocity and head constraint. A recent experimental study [3] shows that the initial CS curvature influences the types of injury and loading observed, indicating its significance. While previous studies [4-6] have developed finite element (FE) models of the human head and neck to assess the sensitivity of CS response to boundary conditions, such as impact plate stiffness and plate angle under HFI, the effects of the CS initial posture have not been assessed. In the current study, a biofidelic head-neck FE model was used to quantify the effect of CS curvature on the predicted neck force and hard tissue injury for HFI. Then, the predicted neck force was compared to HFI data from an experimental study [7].

II. METHODS

A young (26-year-old (yo)) average stature male FE model (Global Human Body Models Consortium (GHBMC) M50-O v5.0) was used in this study. The CS posture predictor (CSP) [8], based on the UMTRI database [9], was used to define target postures using anthropometric inputs (1846 mm of stature and 26yo) and varying the head angle. The head angle was based on the tragion-eye centre angle (TECA), which was obtained from reported root mean square errors associated with regression models used to define the tragion and eye centre locations [10]. Subsequently, the following neck postures (TECA) were generated: 0°, 12.3° and 18.0°. The GHBMC FE model TECA was found to be 5.0°. The resultant four CSP outputs, based on the sagittal plane, were used to obtain the required vertebral landmarks [11], which were then input into the reposturing software (PIPER 1.0.2, PIPER Project, EU). Following the reposturing, PIPER was used to achieve mesh quality similar to the original GHBMC FE model [12]. Moreover, following the specimen preparation outlined in [7], the head and neck regions were extracted; specifically, the ligamentous spine attached to the deformable skull containing the brain (Fig. 1). Then, the four FE models were subjected to a vertex impact, with a head velocity of 3.2 m/s, contacting a rigid plate with a Teflon coating. The T1 vertebra was modified to have a mass of 15 kg to account for the effective torso mass. The conditions applied to the models were based on the experimental setup outlined in [7]. From the FE simulations, the neck axial force history and vertebral hard tissue fracture location and type were monitored [13].

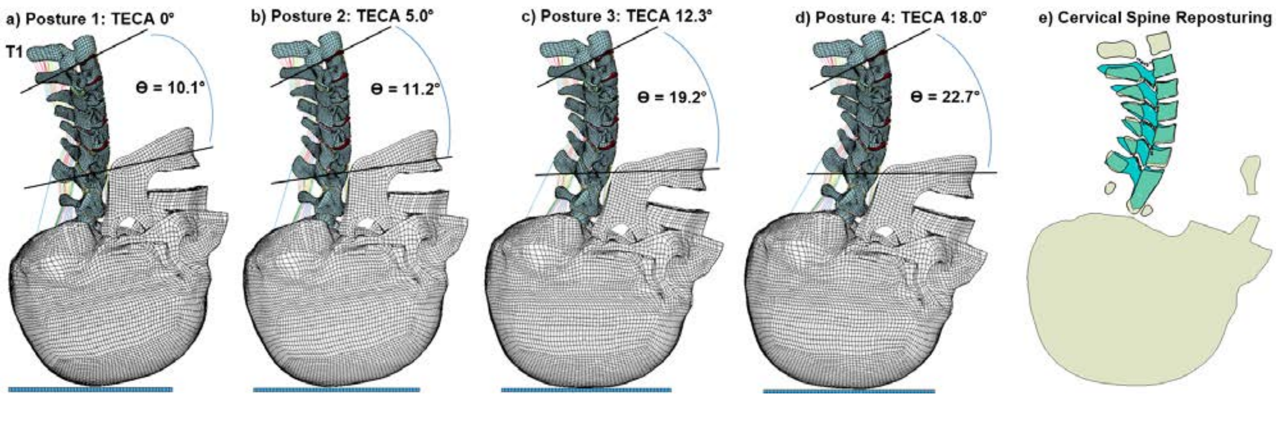

Fig. 1. a) Neck posture 1: TECA=0° and Cobb angle (ϴ)=10.1°; b) Neck posture 2: TECA=5.0° and ϴ=11.2°; c) Neck posture 3: TECA=12.3° and ϴ=19.2°; d) Neck posture 4: TECA=18.0° and ϴ=22.7°; e) GHBMC C-Spine (yellow transparent) and CSP spine (blue).

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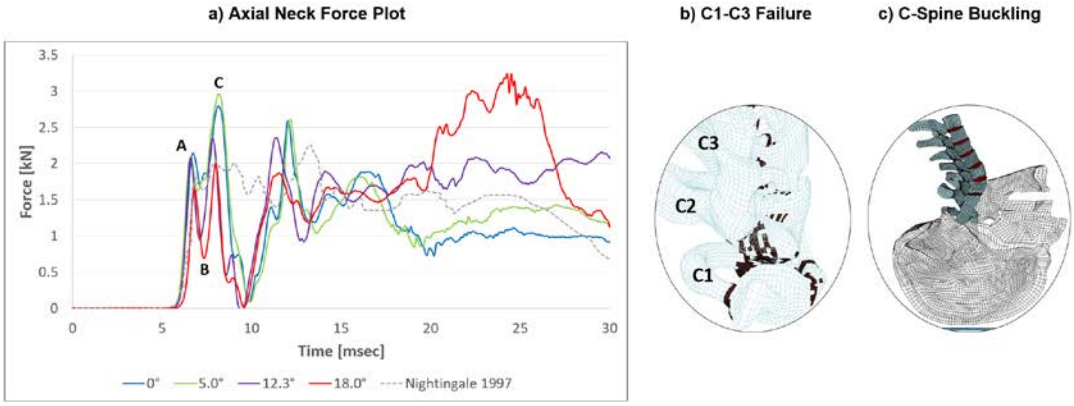
III. INITIAL FINDINGS

As the FE models undergo HFI, the initial head contact with the plate leads to a rise in axial neck force, measured at C7, until the first peak (region A), followed by an axial force drop until the first minimum (region B), and another axial force rise until the second peak (region C) (Fig. 2a)). The rise in axial force until region A after initial head contact was due to the compressive momentum of the following torso (i.e. T1). At region A, the head began to rebound off the plate and the upper CS (C1-C2) experienced flexion, which decreased the axial neck force until region B. The neck axial loading increased after region B due to the CS being compressed by the momentum of the following torso and the rebounding head. Subsequently, the force increased until region C, at which point the CS experienced first order buckling (Fig. 2c)), which alleviated the axial loading on C7 that then led to another force drop. For all postures, failure initiated at C1 anterior and posterior arches, C2 pedicles, and C3 articular pillars between regions A and B (Fig. 2b)). This CS behaviour was observed in all four models. However, with increasing curvature, the axial force corresponding to the first peak decreased (Fig. 2d)).



d) Peak Force Summary

Axial Force at 1st Peak

Posture [°]	Force [kN]
0	2.14
5.0	2.09
12.3	2.08
18.0	1.66

Fig. 2. a) Model vs experiment axial force history; b) C1-C3 failure (black elements); c) C-spine 1st mode buckling; d) Axial force at 1st peak summary table.

IV. DISCUSSION

The drop in load due to increasing neck curvature can be attributed to the fact that the vertebrae were subjected to less compressive force by buckling in first order mode and evading the axial load path. Similarly, for the CS buckling behaviour in region C (Fig. 2a) and c)), the C7 x-velocity in the anterior direction and the C4-C6 extension rotational velocity increased, which collectively reduced the axial stiffness of the CS. The failure observed in the model, which initiated 2.5 ms after initial head contact (Fig. 2b)), had a comparable fracture pattern and onset with the experimental results from Nightingale *et al*. [7], who reported C1 2-part posterior arch fracture and C2 pedicles fracture 2.2 ms after initial head contact. However, in the higher curvature simulations (i.e. postures 3 and 4), there was more middle CS and less upper CS failure observed. Moreover, as the curvature increased from 0° to 18.0° the first peak axial load dropped from 2.14 kN to 1.66 kN (25.3%). The drop in axial load aligns with what is described in the experimental study [3], which reports that CS's with higher curvature are less prone to bony injuries and are associated with lower axial forces. Although increasing neck curvature did not produce a large drop in CS axial load, it had a notable effect on vertebral fracture patterns. Future work will focus on characterising the CS behaviour under HFI in the context of factors such as head rebound, and subsequently determining the influence on kinetic and kinematic response.

V. ACKNOWLEDGEMENTS

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VI. REFERENCES

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