

Effect of Aged Neck Posture on Kinematics and Tissue-Level Response in Frontal Impact Conditions

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I. INTRODUCTION

It has been reported that the elderly population has an increased incidence of injury [1], potentially associated with changes in posture and material properties corresponding to the aging process. Specifically, the incidence of hard tissue fracture increases in the mid cervical spine for ages 65–75 years [2]. A previous study [3] developed an aged occupant finite element (FE) human body model (HBM) focused on aged material properties and morphological changes in the head, rib cage and lower extremities; but with a young cervical spine posture and material properties. Increased lordosis in the cervical spine has been correlated with age [4] and quantified in a cervical spine posture predictor (CSP) database [5]. HBMs are developed in specific positions (e.g. seated or standing) and are commonly repositioned or re-postured using FE simulation or morphing software [6]. Representing an aged cervical spine curvature requires precise bone positioning, which can be challenging to achieve when the morphing software landmarks and database reference points differ. In the current study, a method to re-posture a neck FE model from a young (26-year-old (yo)) to an aged (75yo) posture and retain mesh quality is proposed, and the effect of re-posturing assessed in frontal impact scenarios.

II. METHODS

The head and neck regions were extracted from the current Global Human Body Models Consortium (GHBMC) average stature seated young male FE model (M50-O v4.5) (Fig. 1a). First, the young posture of the GHBMC model was assessed using the CSP with the anthropometrics corresponding to the 26yo model (1,749 mm stature, 0.53 for the standing/seated height ratio). It was found that the subject-specific FE model neck was longer than the population average reported in the CSP by 10.4%. To address this regional difference, the stature in the CSP was increased to 1,846 mm (5.5% increase) to match the FE model young neck length (Fig. 1b). Secondly, the age in the CSP was increased from 26yo to 75yo (Fig. 1c). The resulting CSP data, referenced to the sagittal plane, were used to accurately position the three-dimensional vertebra in a solid modeling package (CATIA V5, Dassault systems, France). The required vertebral landmarks (transverse processes and superior endplate) needed for the repositioning software (PIPER 1.0.2, PIPER Project, EU) were then extracted and applied to age the 26yo FE model to a 75yo FE model.

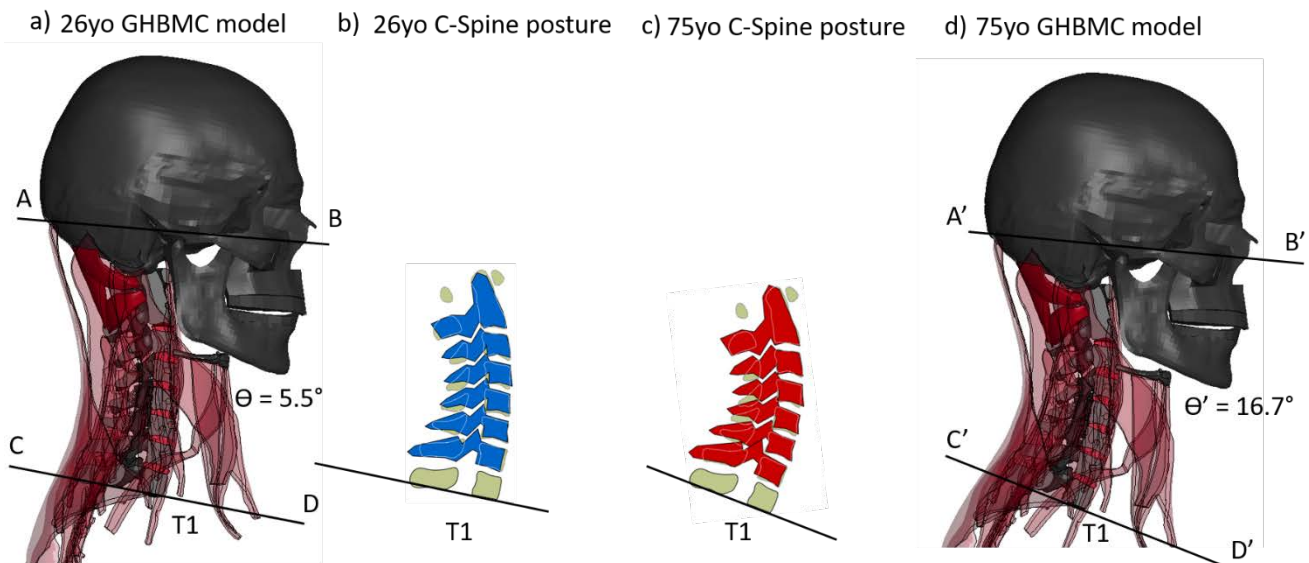


Fig. 1. a) GHBMC young neck model and young Cobb angle (Θ) AB:CD; b) GHBMC young C-Spine curvature (yellow transparent) and 26yo C-Spine prediction (blue); c) GHBMC aged C-Spine curvature (yellow transparent) and 75yo C-Spine prediction (red); d) Aged posture GHBMC neck model and aged Cobb angle (Θ') A'B':C'D'.

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Following the software re-posturing, the soft tissues of the aged posture model were identified to have poor mesh quality (16.7% of the elements) relative to the mesh requirements (warpage < 50°, aspect ratio < 8, skew < 70° and Jacobian > 0.4). The soft tissue mesh of the aged posture model was improved using transformation smoothing tools [7]. The resulting FE mesh (Fig. 1d) met the mesh quality requirements. The aged posture model was then assessed for 8 g and 15 g frontal impact scenarios, and compared to the young posture model using global metrics (head centre of gravity (CG) kinematics) and tissue-level responses (capsular ligament (CL) distraction, intervertebral disc (IVD) strain, superior endplate stress, and relative vertebral kinematics) as exemplary metrics (Fig. 2).

III. INITIAL FINDINGS

Following re-posturing, the IVD height in the aged posture model increased anteriorly (5%) and decreased posteriorly (9%) resulting from translation and rotation of the hard tissues according to the CSP. In addition, the interfacet gap was reduced by 37% on average for all vertebral levels due to re-posturing. In the frontal impact (Fig. 2), the aged posture demonstrated similar resultant linear acceleration of the head CG compared to the young posture, assessed using cross-correlation (15 g_{CORA}=0.97, 8 g_{CORA}=0.95), where a value of 0 indicates no similarity and a value of 1.0 indicates strong similarity.

The aged posture predicted higher anterior intervertebral disc strains in the 8 g (16%) and 15 g (12%) impacts. The stress in the superior endplate in the anterior region of the vertebral bodies increased at all vertebral levels for both impact severities (average of 51% in the 8 g impact and 50% in the 15 g impact). In general, the capsular ligament strain was lower for the aged posture at all cervical spine levels and both impact severities.

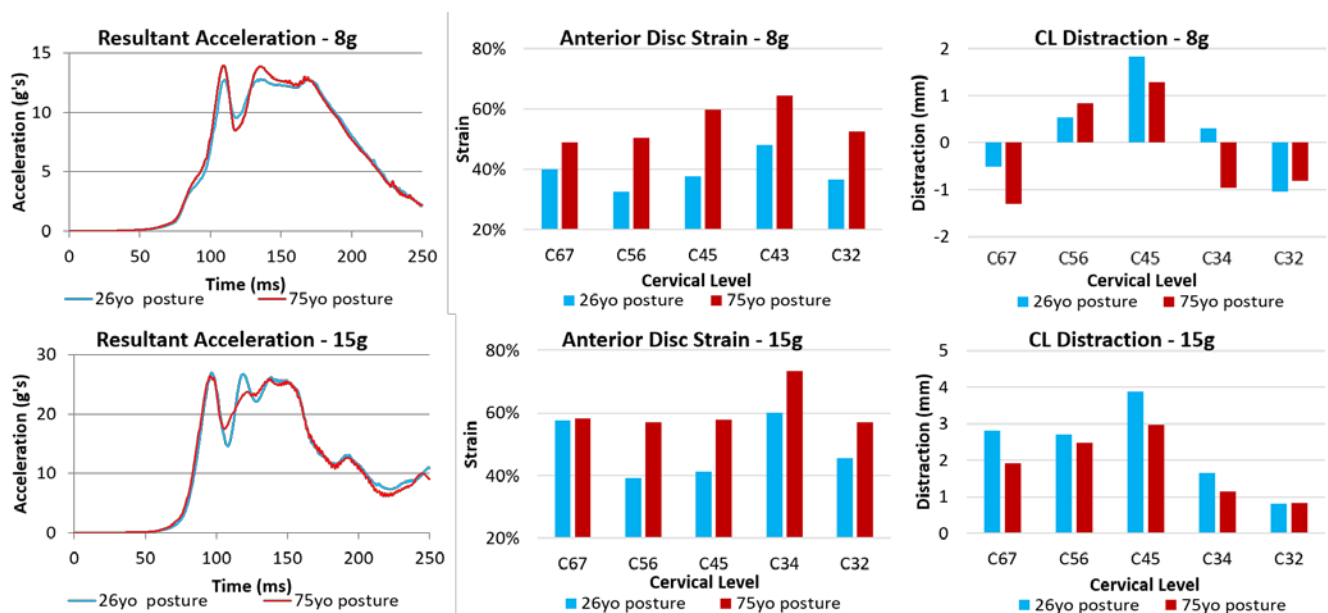


Fig. 2. Comparison of global and local metrics between young and aged postures.

IV. DISCUSSION

The relative vertebral kinematics (translation and rotation of the vertebra) were similar for both postures, leading to similar head kinematics. In contrast, higher disc strains were observed in the aged posture model and were attributed to the anterior disc height reduction, as reported in the CSP database for the aged posture. In the model, higher disc strains were associated with higher stresses in the vertebra at the anterior location. Epidemiological studies have demonstrated 65–75yo subjects exhibit higher levels of hard tissue failure, primarily anterior compression fracture in the lower cervical spine [2], which may be related to the increased stresses in the vertebra. The difference in CL distraction was primarily attributed to a reduced interfacet gap in the aged posture compared to the young posture; however, there is limited data regarding change in the facet gap with age and further investigation is required. In this study, changes in posture from aging resulted in higher tissue-level responses. Future research will consider aged material properties and additional impact scenarios (lateral, rear), which may further amplify the effect of posture on the response.

V. REFERENCES

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