Anatomical Enhancements to the Upper Cervical Spine FE Model Improved the Intervertebral Response Required for Injury Assessments in Non-neutral Neck Postures

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I. INTRODUCTION

Experimental studies have shown an increased risk of neck injury among car occupants with a non-neutral neck posture [1]. Detailed finite element (FE) neck models (NMs), originally developed in neutral postures, could provide valuable insight into injury risk for non-neutral postures, but require biofidelic repositioning. Often, FE NMs are repositioned to a non-neutral posture within the physiologic range of motion (ROM) through stress-free morphing while a recent study demonstrated the importance of retaining the initial strains in repositioned FE NMs [2]. Therefore, it becomes important to ensure accurate vertebra positions for repositioning in the physiologic regime to predict tissue-level strains. The upper cervical spine (UCS) is an important contributor to physiologic neck motions [3-4]. Specifically, the C0-C1 primarily contributes to sagittal motion, while C1-C2 contributes to axial rotation [3-4]. Existing FE NMs; however, have assessed the UCS as one segment in a dynamic regime [5], while it is important to assess the physiological ROM of the UCS at the intervertebral levels. The current study identified the need for enhancements to the geometry of a contemporary UCS FE model and then quantified the outcomes using independent experimental data.

II. METHODS

The UCS model (UCS_{M50}), previously validated for C0-C2 kinematics [5], was extracted from the Global Human Body Model Consortium (GHBMC M50 v.4.5). The C2 was constrained, and a moment of 1.5 Nm was applied to C0 in flexion and extension to simulate the experimental set-up [3]. The model was solved using commercial finite element software (LS-DYNA R9.2). The intervertebral kinematic responses in flexion and extension from 0 Nm to 1.5 Nm were assessed, and a need for geometrical enhancements to the UCS_{M50} was identified, including the C0-C1 joint space and alar ligament insertion points, to be consistent with the literature. The C0-C1 joint space in the UCS_{M50} was 0.2 mm, relatively small compared to the reported average value of 1.0 mm with a range of 0.5 mm to 1.8 mm [6]. The joint space was increased to 0.5 mm ($UCS_{M50}|JS$), the lower bound reported in the literature, by decreasing the C1 cartilage thickness. Decreasing the C1 cartilage thickness further, to increase the joint space beyond 0.5 mm would have led to significant re-meshing of the model. The alar ligaments that were partially inserted on the skull in the UCS_{M50} were reoriented to have all the insertion points on the medial occipital-condyle (OC) ($UCS_{M50}|JS|alar$), as reported in the literature [7] (Fig. 1a). The sagittal intervertebral kinematic responses from the UCS_{M50}, $UCS_{M50}|JS$ and $UCS_{M50}|JS|alar$ models were compared with the experimental data [3]. For model assessments, the absolute percentage difference between the model response and experimental data was calculated. The model was assessed at the C0-C1 and C1-C2 intervertebral levels, across 0 Nm to 1.5 Nm at 0.5 Nm load increments, in flexion and extension. Further, the average of the absolute percentage difference between the model and the experimental data was calculated across 0 Nm to 1.5 Nm individually for C0-C1 flexion, C0-C1 extension, C1-C2 flexion and C1-C2 extension.

III. INITIAL FINDINGS

The average absolute percentage difference between the experimental data [3] and UCS_{M50} was 63% in C0-C1 flexion, 70% in C0-C1 extension, 75% in C1-C2 flexion and 31% in C1-C2 extension. The response in the $UCS_{M50}|JS$ model improved as the average absolute percentage difference reduced to 17% in C0-C1 flexion, 19% in C0-C1 extension, 54% in C1-C2 flexion and 25% in C1-C2 extension relative to [3]. The average absolute percentage difference between $UCS_{M50}|JS|alar$ model and [3] was 20% in C0-C1 flexion, 19% in C0-C1 extension, 55% in C1-C2 flexion and 36% in C1-C2 extension, representing an overall improvement in the model response.

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The $UCS_{M50}|JS$ model average response increased by 9.5° in C0-C1 flexion, 7.3° in C0-C1 extension and 2.5° in C1-C2 flexion, while there was a subtle change in C1-C2 extension, when compared with the UCS_{M50} model. The $UCS_{M50}|JS|alar$ model average response increased by 1° in C0-C1 flexion and C1-C2 extension, 3° in C0-C1 extension and reduced by 0.1° in C1-C2 flexion when compared with the $UCS_{M50}|JS$ model (Fig. 2).

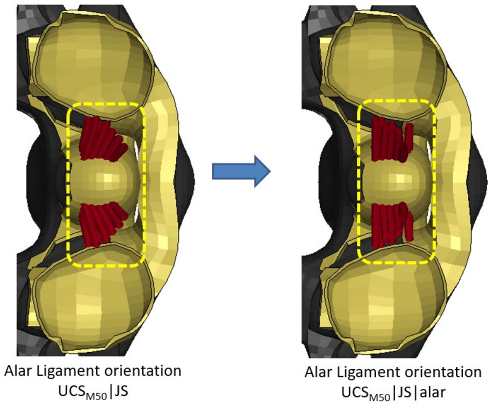

Fig. 1a. Improved alar ligament insertion based on [7]

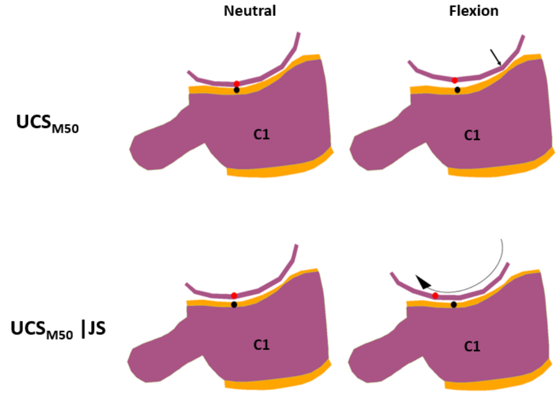

Fig. 1b. Relative motion of C0 over C1 in flexion

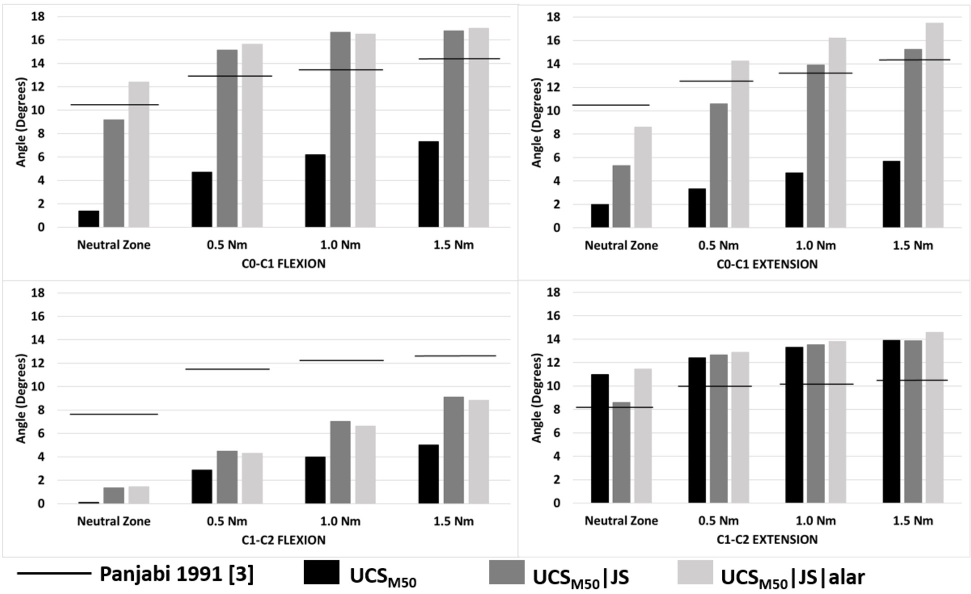

Fig.2. Intervertebral responses of the original and geometrical enhanced UCS models.

IV. DISCUSSION

An initial assessment of the UCS_{M50} model at the C0-C1 joint showed the OC lifting up rather than rolling and sliding over the C1 cartilage, as reported in the literature [7]. The unphysical C0-C1 joint motion was attributed, in part, to the low joint space. Increasing the C0-C1 joint space allowed the OC to roll and slide over the C1 cartilage surface in flexion and extension loading modes but had relatively lesser effects on the C1-C2 kinematics (Fig. 1b). Correcting the alar ligament orientation in the $UCS_{M50}|JS$ increased the C0-C1 extension (average of 3° increase) and had a subtle influence on the other cases (Fig. 2). The study demonstrates that small anatomical enhancements could have an important effect on the model response. The UCS model was evaluated at the intervertebral level rather than the whole UCS level to improve the UCS intervertebral kinematics and therefore improve tissue strain predictions in FE NM repositioning and assessment of out-of-position impact investigations.

V. REFERENCES

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