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EVALUATION OF OPTIMISED CERVICAL SPINE VISCOELASTIC ELEMENTS FOR SPORT INJURY ANALYSIS

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INTRODUCTION

Musculoskeletal models of the cervical spine are valuable tools to assess the intervertebral loads and injury mechanisms of sporting head impacts and provide an intermediate step between in-vivo experimental trials and detailed finite element analysis. A previous musculoskeletal model created for estimating intervertebral joint loading during rugby contact [1] events included kinematic constraints, which neglected individual joint translation and therefore may oversimplify the behavior of the cervical spine during impacts. In order to both confidently estimate the internal loading and resulting vertebral kinematics, intervertebral viscoelastic ("bushing") elements should be included in the model. Such viscoelastic elements are used to describe the behaviour of the joints' passive structures [2] and allow for resulting motions to be calculated compared to kinematic constraints. A key challenge is to identify the optimal bushing element that can generate reliable kinematics of the model's cervical spine with respect to the load applied. Therefore, the aim of this study was to assess the performance of anteroposterior shear viscoelastic elements, that were validated together with compressive elements for impulsive axial loading conditions, under loads containing shear components.

METHODS

Axial and anteroposterior viscoelastic bushing parameters were estimated via an optimisation procedure in a combined in-vitro and in-silico study. The optimisation procedure that identified the parameters included forward dynamic simulations of five specimen specific models that were driven by the experimental axial loads in order to replicate the in-vitro experiment. We carried out a 1000 sample Monte-Carlo analysis that perturbed the viscoelastic parameters by ±50% from their optimised values to evaluate the models' sensitivity to a range of parameter values under the same axial loads. This method classified which parameters had the greatest effect on the models' performance. The initial parameters estimated by the optimisation procedure were then implemented in the C2-C3, C3-C4, C4-C5 and C5-C6 joints of the population specific musculoskeletal model [1]. Forward dynamic simulations were executed with the updated musculoskeletal model of the cervical spine and head positioned parallel to the horizontal. An initial simulation with a constant anterior shear load of 250 N applied to the base of the head was completed to compare the model's anteroposterior joint displacement against available experimental data. A second simulation investigated a possible injurious loading scenario that could be experienced in a head first impact in rugby. Experimental loads were collected from an instrumented ATD headform during live scrummaging trials. The compression flexion load was scaled to included maximum values of 1100 N and 700 N for anterior shear and compression respectively. A final simulation applied impulses with higher maximal values of 2500 N and 1500 N for shear and compressive forces respectively after the experimental load was scaled further. For all simulations, peak values of the applied loads were reached in 60 ms after an initial 40 ms latent period. No muscle activation was prescribed to reduce the active muscle force effects but their passive contribution remained.

RESULTS AND DISCUSSION

The values of the bushing parameters identified by the optimisation increased from their respective initialised values and were 23.1 MN/m for axial stiffness, 4300 Ns/m for axial damping, 75.9 kN/m for shear stiffness and 1400 Ns/m for shear damping. Perturbations in shear damping and axial stiffness had the most effect on model performance. Decreasing values of shear damping resulted in larger error in the sensitivity analysis (Figure 1). Axial stiffness also showed an effect of increasing or decreasing values however this effect was smaller than the one observed in shear damping. This is possibly caused by the considerably higher values of axial stiffness (23.1 MN/m) compared to shear damping (1400 Ns/m), which, even when perturbed by – 50 %, remain in the same order of magnitude. On the contrary, the same proportional perturbation of the optimised shear damping reduces the value by an order of magnitude.

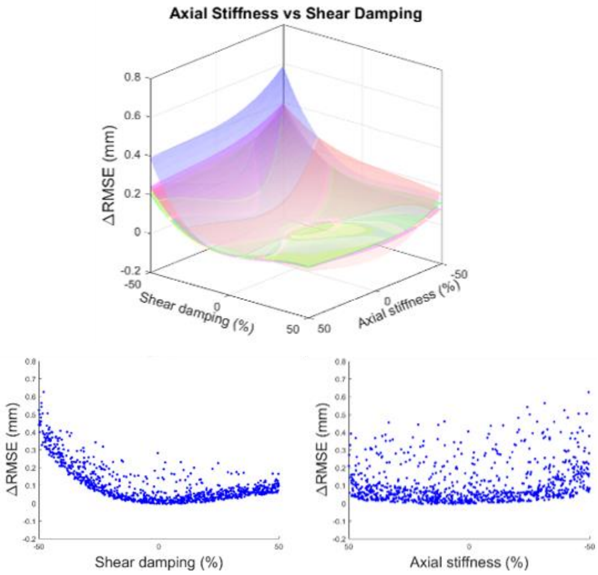

Figure 1. Axonometric view shows the response of five optimised models as the interpolated 3rd degree polynomial surfaces (above). Projections of each axis of the parameter perturbation against the calculated tracking error (ΔRMSE) for one model as an example (below).

During these impulsive axial loads applied to the cervical spines in the experiments a buckling response was observed as previously observed by [3]. The anterior shear motion of the vertebrae caused by the loading of the specimens in the current study, however, did not lead to injuries because the applied load was chosen to be sub-catastrophic. The sensitivity analysis thus highlighted the importance of correct choice for anteroposterior joint damping parameters used in musculoskeletal model of the cervical as lower values chosen for axial impacts may result in misrepresentative kinematic responses of the model cervical joints in anteroposterior translation.

The three forward dynamic simulations resulted in anterior cervical spine joint displacements representative of the loads applied to the base of the head segment (Figure 2). Anterior displacements of the C2-C3, C3-C4, C4-C5 and C5-C6 joints ranged from 3 to 3.5 mm under the 250 N pure shear load applied to the base of the head. These values are in line with the 2 mm anterior joint displacements of three-segment specimens produced under non-catastrophic quasi-static experimental loads of 220 N recorded by [4]. During the second impact simulation, the cervical joints displaced anteriorly by 9 to 10 mm. The simulation that reached 2500 N anterior shear load resulted in maximal anterior displacements of 13 to 19 mm.

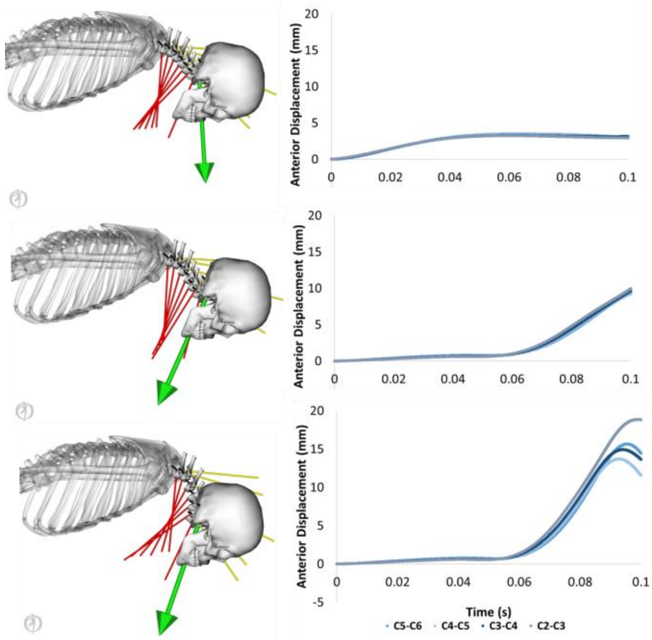

Figure 2. Anterior cervical spine joint displacement as a result of the three applied loads (green arrows). The kinematic results of the individual joints (right column) and final model pose (left column) are presented for the static 250 N load (top row), 1100 N shear and 700 N compressive loads (middle row) and 2500 N shear and 1500 N compressive loads (bottom row).

Anterior shear failure tests on porcine cervical specimens have shown maximal displacements of 16 to 22 mm under loads of 2300 to 3500 N [5]. It should be noted that both studies used ex-vivo specimens with musculature removed, which is a simplification of the complex mechanism of the cervical spine. Also, the clinical presentation of cervical spine injuries sustained in-vivo by similar loading patterns are heterogeneous [6], making the identification of injury mechanism challenging.

CONCLUSIONS

Cervical spine musculoskeletal models used in the analysis of axial impacts are sensitive to anteroposterior joint shear damping parameter values and that shear values estimated under such axial impacts provide a viable approach to analyse loads representative of sport collisions.

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