Effects of Controlled Muscle Activations on Human Head–Neck Responses during Low-Speed Rear Impacts

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I. INTRODUCTION

Approximately 50% of all injuries caused by traffic accidents are neck injuries, and rear-end accidents account for approximately 60% of neck injuries in Japan. The most frequent injury is whiplash injury accompanied by neck pain due to low-speed rear impacts. However, the mechanism of such a neck injury is still unknown. One hypothesis, proposed in previous studies based on post-mortem human subject (PMHS) tests and volunteer tests, is that the injury is related to the relative motion between the head and torso. A useful method for evaluating the proposed hypothesis is computer simulation using finite element (FE) models of the human body. Previous studies have revealed that active muscle models are effective for investigating the injury mechanism; however, only a few analyses considering active muscle models have been performed. The objective of this study is to evaluate an activation controller and its resulting effects by using an FE model of the human body with active muscles. In this study, a newly developed activation controller for muscles was applied to the model by using newly added functions of LS-DYNA. By using this model, it was possible to evaluate the head–neck responses in low-speed rear-impact situations.

II. METHODS

Simulation analyses of an FE model of the human body with muscles activated by an activation controller were performed by using a nonlinear explicit FE solver, LS-DYNA v971 R8.0.0 (LSTC, USA).

Head–Neck Model with Muscles

An FE model of the human body, THUMS v5 [1], was used in this study. Fig. 1 shows the head–neck region of THUMS. Muscle parts were modelled using Hill-type muscle characteristics, while each muscle had the capability of generating forces based on given activation levels.

Simulation setups for rear impacts were designed based on experimental studies conducted by Ono *et al.* [2] and White *et al.* [3]. The selected impact speed for those analyses was 8 km/h. Enforced motions were given to the first thoracic vertebra (T1); specifically, the motions were defined as translational velocity in the X- and Z-directions and a rotational angle around the Y-axis in a global coordinate system. For simplification, body parts lower than T1 followed the kinematics of T1. The head rotational angle is defined in a local coordinate system as the change in head angle from the initial posture, as shown in Fig. 2. An uncontrolled head–neck model was represented with 1% activation level of all muscles, and a controlled head–neck model was represented with activation patterns generated by an activation controller as described in the next section.

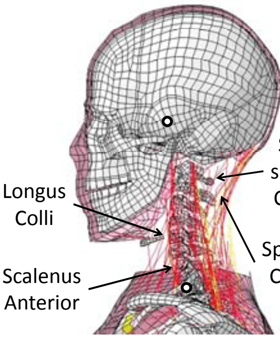

Fig. 1. Head-neck model with muscles.

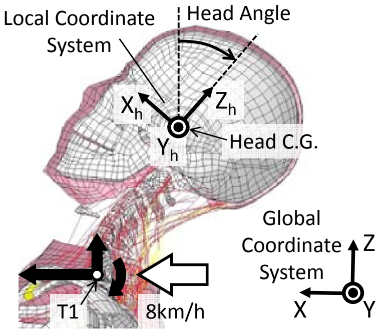

Fig. 2. Definition of head rotational angle.

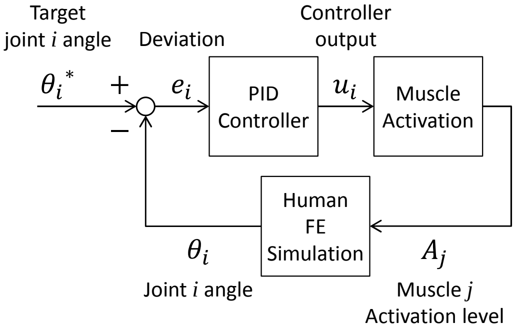

Fig. 3. Control system of muscle activation.

Muscle Activation Control System

Twenty-three neck muscles exist on one side of the model. As a representation, Fig. 1 indicates four muscles. The Longus Colli and the Scalenus Anterior are flexor muscles in the head–neck region, whereas the

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Semispinalis Capitis and the Splenius Capitis are extensor muscles. Fig. 3 shows the applied muscle activation control system, which is constructed as a feedback system. The activation levels of the muscles were calculated based on the difference between the target joint angle and the current joint angle. In this study, the current joint angle was set as the head rotational angle, and the target joint angle was set as the initial head rotational angle to maintain the posture. The system mainly consisted of proportional-integral-derivative (PID) controllers and a muscle activation component. The PID controllers were represented by the PIDCTL function, which was recently introduced into LS-DYNA v971 R8.0.0. The muscle activation component calculates the activation level of each muscle based on the activation ratios between the flexor and extensor muscles and the percentage contributions of the flexor and extensor muscles for head–neck motions. In this study, delay time was not considered for simplicity and because of a lack of data.

III. RESULTS

Fig. 4 shows the comparison of head rotational angles obtained from simulation results and previous experimental results [2–3]. Comparing the PMHS test and the uncontrolled model, the model shows similar behaviour to the experimental data until arriving at the maximum rotational angle. Additionally, the maximum angle of the controlled model was less than that of the uncontrolled model. This change is similar to the relation between the results of the PMHS test and the volunteer test. This is caused by the tendency of the activated muscles to maintain the head rotational angle in the initial posture. Fig. 5 shows the muscle activation levels of the representative muscles obtained from the simulation results of the controlled model. As the head rotated backward during rear impact, the flexor muscles were activated more than the extensor muscles.

However, the simulation results did not show good agreement with the test results after 0.23 s. In the simulation results of the uncontrolled model, the head tended to return to the initial angle earlier, compared to the PMHS tests. In order to obtain a solution to this problem, more studies are required on modelling a cervical spine in THUMS. In addition, a parameter study or review of the activation control system is required in the future. Even if simulation results of the head rotational angle agree well with the experimental data, it is not known whether the muscle activation levels are similar in an actual human body. Therefore, verification of the activation levels with experimental data, such as measured electromyography, is necessary.

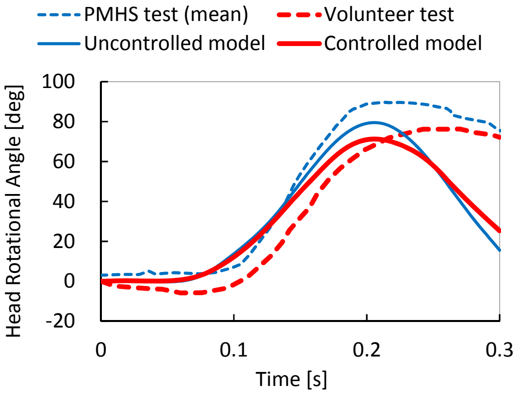

Fig. 4. Comparison of head rotational angles.

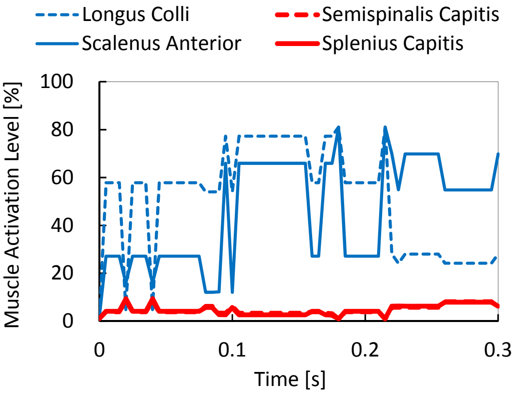

Fig. 5. Muscle activation levels produced by the control system.

IV. CONCLUSIONS

A control system for muscle activation was applied to an FE model of a human body (THUMS v5). The system was represented by the newly added function (PIDCTL) of LS-DYNA. From the simulation results of a low-speed rear-impact situation, the activated muscles tended to maintain the head rotational angle toward the initial posture; this tendency is similar to an actual human body. Further studies are required to evaluate the control algorithm for the muscle activations by comparing the activation levels with electromyography measurement data.

V. REFERENCES

[1] Iwamoto, M. *et al.*, *Stapp Car Crash Journal*, 2015. [2] Ono, K. *et al.*, *Proc. Stapp Car Crash Conf.*, 1997.
[3] White, N. A. *et al.*, *SAE Technical Paper*, 2009.