Effect of Different Muscle Fibre Types on the Neck Kinematics for Frontal Impact

Bartłomiej M. Pilarczyk, Ciaran Simms, Duane S. Cronin

I. INTRODUCTION

Skeletal muscle in the human body is composed of two main fibre types: type I (slow twitch) and type II (fast twitch) fibres. Type I is suited for functions in which moderate forces are required for prolonged periods, e.g., postural maintenance, while type II fibres are capable of short bursts of intense activity, e.g., fast running, weight lifting [2][7]. Type II fibres are also active in the neck musculature during a car impact. During an impact, stronger muscles within the neck complex provide protection for the upper cervical spine [5], attributed to type II muscle fibres. However, the size of type II fibres decreases with age [7-10]. Therefore, it is important to assess whether different fibre composition of skeletal muscle in the human neck may have an influence on the head kinematics. The objective of this study was to evaluate changes in head kinematic response due to variations in muscle fibre composition.

II. METHODS

A neck and head finite element model (Global Human Body Models Consortium (GHBMC) 50th percentile male, M50 ver. 4.3) [6] (Fig. 1) validated with intermediate and high severity impacts was used. This model includes the cervical vertebrae (C1-C7), first thoracic vertebra (T1), cervical spine ligaments, intervertebral discs, 27 pairs of cervical muscles and the skin. Passive muscle behaviour is modelled by three-dimensional solid elements with a hyperelastic response, while active muscle behaviour is modelled using one-dimensional Hill-type active elements. The stress developed by a single active muscle element is a function of four parameters: a(t), muscle activation level (neural excitation); σ_{max}, maximal isometric stress; $f_L(L)$, force-length relation and $f_V(V)$, force-velocity relation (1). The force-velocity curve $f_V(V)$ (2) depends on the fibre type properties (Table I) and has a strong influence on the resulting muscle force [1-3]. The $f_V(V)$ curve was defined to represent two different fibre types (slow and fast twitch) [2-3] and a mixed type fibre model. The shapes of the curves differ, particularly for V > 0 (Fig. 2). Consequently, three different $f_V(V)$ curves were used in the simulation of the frontal impact: slow twitch, fast twitch and a mixed type model.

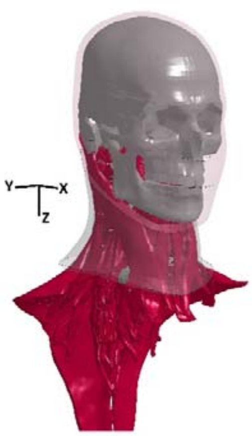

Fig. 1 GHBMC M50: Neck-Head complex.

$$\sigma_{CE} = a(t) \times \sigma_{max} \times f_L(L) \times f_V(V) \quad (1)$$

$$f_V(V) = \begin{cases} 0 & V \le -1 \\ \dfrac{1+V}{1-\dfrac{V}{C_{sh}}} & -1 < V \le 0 \\ \dfrac{1+V\dfrac{C_{ml}}{C_{leng}}}{1+\dfrac{V}{C_{leng}}} & V > 0 \end{cases} \quad (2)$$

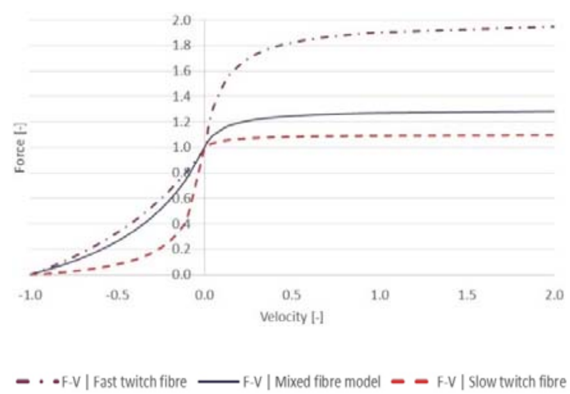

Fig. 2 F-V relations for different fibre types.

TABLE I
MUSCLE FIBRES PARAMETERS[2][4]

Parameter / Fibre type	Slow fibres	Mixed fibre model	Fast fibres
V_{max}	$2l_0/s$	$5l_0/s$	$8l_0/s$
C_{sh}	0.1	0.55	1
C_{ml}	1.1	1.3	2

l_0 - muscle length at rest
V_{max} - maximum contraction velocity
C_{sh} - determines shape for concentric contraction
C_{ml} - determines shape for eccentric contraction
C_{leng} - determines transition between eccentric and concentric contraction (here: 0.1065) [4]

The specific ratio of fibre types present in the skeletal muscle is challenging to assess, therefore the muscle

BM. Pilarczyk (bpilarczyk@uwaterloo.ca, +1-519-888-4567x38467) is a PhD candidate and DS. Cronin (duane.cronin@uwaterloo.ca) is a Professor in the Department of Mechanical and Mechatronics Engineering, University of Waterloo, Canada. C. Simms (csimms@tcd.ie) is a Professor in the Department of Mechanical and Manufacturing Engineering in Trinity College, Dublin, Ireland.

model implemented in the current neck model uses the mixed properties of two fibre types, and is referred to as *mixed* fibre model. The mixed fibre model assumes equal ratio of the two fibre types in a skeletal muscle [4].

Simulations on a single muscle element were performed in order to evaluate the difference in response between the two fibre types and the mixed fibre model (Fig. 3). The simulation boundary conditions included: activation set to 1 ($a(t) = 1.0$), maximum isometric stress: $\sigma_{max} = 0.5\ [MPa]$ [2][4], and the $f_L(L)$ curve defined as in [4].

For the full neck simulations, two levels of impact severity were investigated: (1) peak acceleration of ~6G (impact velocity: 30-40 km/h) and (2) peak acceleration of ~14G (impact velocity: >55 km/h). Boundary conditions for the impact simulations are based on the experimental data reported by the Naval BioDynamics Laboratory (NBDL) [11].

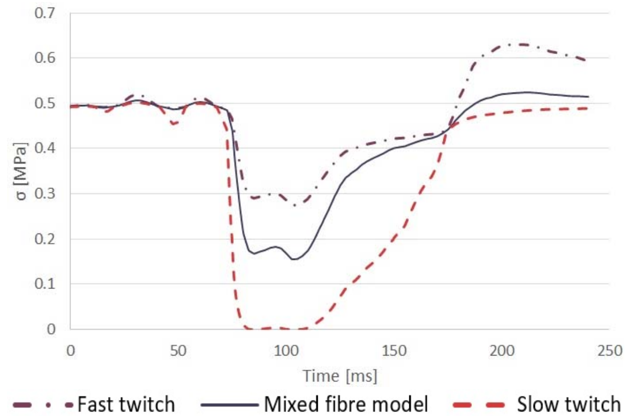

Fig. 3. Resultant stress in single muscle model for different $f_V(V)$ curves.

III. INITIAL FINDINGS

For each test case, kinematic response of the head centre of gravity was evaluated including: X-acceleration (AX), X-displacement (DX), Z-acceleration (AZ), Z-displacement (DZ), Y-rotational acceleration (RAY) and Y-rotational displacement (RDY) (axis system shown in Fig. 1).

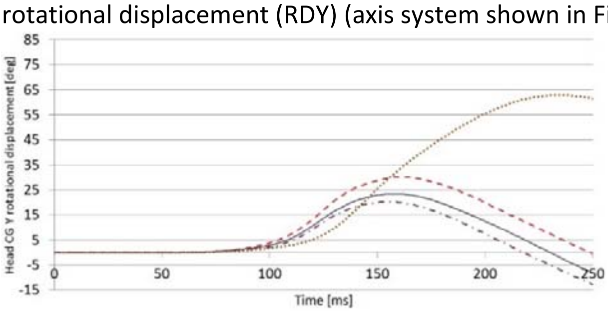

Fig. 4a Head CG: RDY @ ~6G.

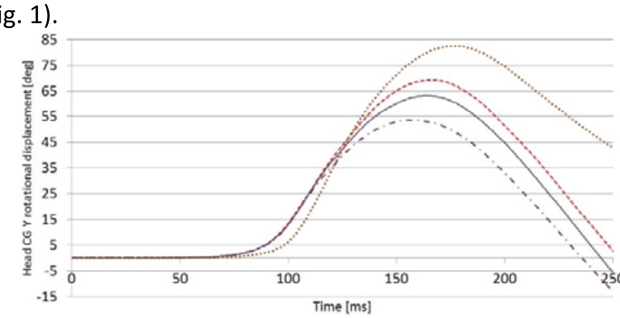

Fig. 5a Head CG: RDY @ ~14G.

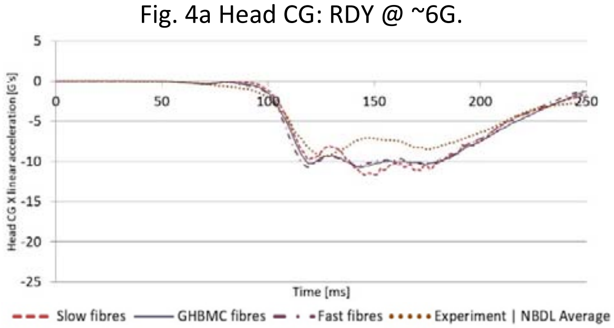

Fig. 4b Head CG: AX @ ~6G.

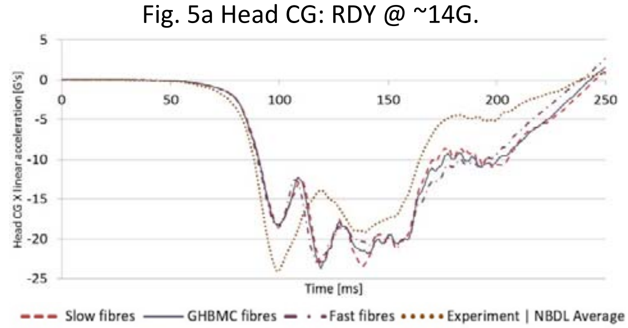

Fig. 5b Head CG: AX @ ~14G.

Rotational displacement (Fig. 4a, 5a) and head CG x-acceleration (Fig. 4b, 5b) followed a similar trend for all cases. Simulations with slow twitch fibres predicted larger head CG forward rotation during the impact compared to the mixed fibre model. Considering only fast twitch fibres, the model predicted a lower head rotation relative to the mixed fibres, where higher rotation may be associated with higher tissue strains and therefore a higher potential for injury to soft and hard tissues, e.g., strain, sprain, or hard tissue injuries. It was noted that the simulation exhibited lower rotation for the low severity impact, compared to the experiments.

IV. DISCUSSION

The kinematic behavior of the head-neck complex demonstrated sensitivity to force-velocity relations defined for different types of muscle tissue fibres, with slow twitch fibres resulting in larger rotation compared to the mixed fibre model. Therefore a reduction of type II muscle fibre size in an aging population (>60 YO) [9] results in increased head kinematics, and corresponding increases in tissue strains, with a higher potential for injury. Moreover, in an older population the muscle activation time may increase. Future investigations should address the combined effect of these two factors on head kinematics, the potential for soft tissue injuries and model response for lower severity impacts.

V. REFERENCES

[1] Wittek et al., IntJOccSafErgo, 1998.
[2] Winters J, MultMuscSys, 1990.
[3] Östh J, PhD thesis, 2014.
[4] Panzer MB, MASc thesis, 2006.
[5] Brolin K et al, Traffic Inj Prev, 2005.
[6] Schwartz D et al, Traffic Inj Prev, 2015.
[7] Standring S, GAnatomy, 2008.
[8] Faulkner J. et al., ClExpPharPhysi, 2007.
[9] Deschness MR, SportsMed, 2004.
[10] Lexell et al., JNS, 1988.
[11] Ewing CL et al., DTIC Document, 1972.