Defining Injury Criteria for the Muscle-Tendon-Unit

Lennart V. Nölle, Syn Schmitt, Oleksandr V. Martynenko

I. INTRODUCTION

Many injury criteria for different parts of the human body have been defined and are currently used for the evaluation of occupant safety in car crash scenarios [1]. Even though significant efforts in this field have been made, accident-related injuries of the muscle-tendon-unit (MTU) have not yet been taken into consideration [2]. Determining the extent of musculature damage is becoming more important, as the repositioning of occupants in highly automated vehicles with technical devices acting against the human body's muscle forces during the pre-crash phase, may need to be accounted for. One of the most prevalent types of direct muscle trauma is strain injury caused by eccentric muscle contraction [3]. Strain injuries can damage both the muscle belly and the connected tendon in various ways, with symptoms ranging from minor functional impairments to the complete mechanical disruption of the strained muscle or tendon [4-5].

Therefore, the goal of this study is to define injury criteria of distinct injury severity in order to evaluate MTU damage during accidents. These injury criteria are presented in relation to MTU behaviour simulated with the arm extracted from the THUMS v5.02 AHBM and the extended Hill-type material (EHTM) implemented and described in [6], as it is the only muscle material in LS-DYNA that includes both the muscle and the tendon models.

II. METHODS

Experimental Data

Three distinct injury thresholds were set analogous to the generic material deformation stages of a standard engineering stress-strain curve. The minor and major MTU injury thresholds were defined as the start of the strain hardening region and the start of the necking region, respectively. The rupture threshold coincides with the failure of the material. Experimental data for all three regions were obtained from literature. Tendon strain injury experiments are available in publication [5]. Here, straining of rat tail tendon fascicles led to functional impairment at 2% strain, structural collagen disruption at 4–6% strain and macroscopic tendon failure at 10% strain. These values correlate with the strain hardening, necking and failure stages of the tendon stress-strain curve given in [5]. A difference in injury behaviour between passive unstimulated and active stimulated skeletal muscle is demonstrated in [7]. Because of this, two different threshold sets for passive and active muscles were determined. Injury data on passive skeletal muscle was taken from [3], where tibialis anterior (TA) and extensor digitorum longus (EDL) muscles of rabbits sustained first signs of injury at 30% of the tensile force needed to pull the muscle to failure (F_{tf}), while study [8] reported the major plastic deformation of rabbit TA muscles at 80% F_{tf}. Publication [4] provided data on the strain injury of maximally activated rabbit TA and EDL muscles. Injury was first detected at 70% F_{tf}, while significant plastic muscle fibre disruption occurred at 90% F_{tf}. A summary of the three injury threshold sets derived from the literature data is given in Table I.

TABLE I
INJURY THRESHOLDS OF TENDON, PASSIVE MUSCLE AND ACTIVE MUSCLE

Type of Injury	Tendon	Passive Muscle	Active Muscle
Minor Injury	2% strain	30% F_{tf}	70% F_{tf}
Major Injury	5% strain	80% F_{tf}	90% F_{tf}
Rupture	10% strain	100% F_{tf}	100% F_{tf}
References	Stauber *et al.* [5]	Noonan *et al.* [3] Nikolaou *et al.* [8]	Hasselman *et al.* [4]

L. V. Nölle (e-mail: lennart.noelle@imsb.uni-stuttgart.de; tel: +49 711 685-68040) is a PhD student, S. Schmitt is a Professor and O. V. Martynenko is a Senior Researcher at the Institute for Modelling and Simulation of Biomechanical Systems, University of Stuttgart.

Computational Modelling

The injury thresholds of the passive and active muscle are derived from the muscle's tensile failure force $F_{tf,}$ which is neither directly defined in the EHTM nor readily available in literature. However, a comparison between the maximum isometric muscle forces F_{max} and the F_{tf} values of EDL muscles measured in [3-4] showed that F_{tf} was 3.32 ± 0.17 times F_{max}. Therefore, F_{tf} was estimated as three times F_{max}, which is given in the EHTM. To account for the different activity levels of the muscle from unstimulated to fully stimulated active, the injury threshold values were linearly interpolated according to Equation 1:

$$F_{thres}(a) = F_{thres,pa} + a\,(F_{thres,ac} - F_{thres,pa}) \tag{1}$$

where F_{thres} is the muscle threshold force (N), $F_{thres,pa}$ is the passive muscle injury threshold (N), $F_{thres,ac}$ is the active muscle injury threshold (N) and a is the activity level of the muscle between 0 and 1.

The upper right extremity model and three of the different muscle control strategies previously described in [9] were used to simulate a simple arm movement, which was compared with the injury threshold sets of both muscle and tendon. The movement comprises an initial stage (t=0 to t=2 s), where only a gravitational load is applied, and a second stage (t=2 s to t=4.5 s), in which the flexor muscles are contracted so that the forearm moves towards the upper arm in a flexion movement. As muscle strain injury occurs during eccentric muscle contraction, only muscle forces and tendon strains of a muscle that is extended during arm flexion are of relevance for the MTU injury evaluation. Consequently, only the results of the monoarticular elbow extensor muscle (MEE) will be shown in the next section.

III. INITIAL FINDINGS

The force-activation and strain-time curves of the MEE for each of the three controllers, as well as the corresponding tendon and muscle injury thresholds, can be seen in Fig. 1 and Fig. 2.

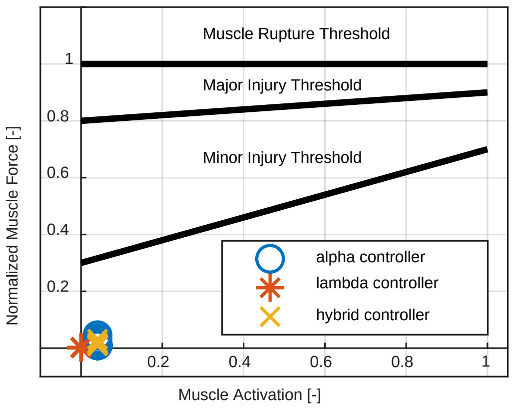

Fig. 1. Force-activation curve of MEE.

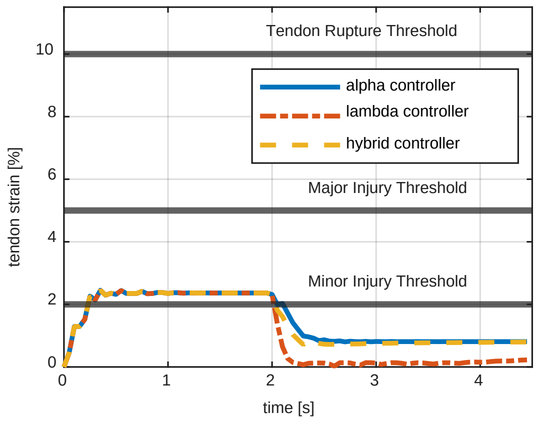

Fig. 2. Strain-time curve of MEE.

IV. DISCUSSION

As expected, a simulation of the physiological movement of the arm does not cause any form of muscle injury (Fig. 1). However, the tendon's strain-time curve in Fig. 2 shows minor tendon injury for all three of the controller types. These high strains during the dropping stage of the arm movement could be explained by the current multi-body model properties used for the simulations. It has an ideal kinematic joint, with no added damping or stiffness of the soft tissues. Thus, forces in the tendon are overestimated for the movement under the gravitational load. The thresholds should therefore be compared to more biofidelic models in future works. Additionally, the tendon injury threshold values may have been set too low, as they were derived from rat tail tendon fascicles and not from tendons found in the human arm. The injury criteria proposed in this study are subject to some limitations. First, other essential types of muscle injury, like lacerations and contusions [10], cannot be modelled with the EHTM as it is a 1-D truss element without a volume that could sustain damage through pressure. Secondly, F_{tf} is estimated empirically and errors in this assumption directly impact the threshold values for muscle injuries. Lastly, viscoelastic material behaviour of muscle was not accounted for during the definition of the MTU injury criteria. Therefore, their application to scenarios in which repeated movements are of interest is not advisable.

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VI. REFERENCES

[1] Goodwin, B. D., *et al*., Springer, 2018.
[2] cahrs.training gmbh, cahrs gmbh, 2020.
[3] Noonan, T. J., *et al*., *AM J Sports Med*, 1994.
[4] Hasselman, C. T., *et al.*, *AM J Sports Med*, 1995.
[5] Stauber, T., *et al*., *Matrix Biol*, 2019.
[6] Kleinbach, C., *et al*., *Biomed Eng Online*, 2017.
[7] Garrett, W. E., *et al*., *AM J Sports Med*, 1987.
[8] Nikolaou, P. K., *et al*., *AM J Sports Med*, 1987.
[9] Wochner, I., *et al*., IRCOBI, 2019.
[10] Baoge, L., *et al*., ISRN Orthop, 2012.