Simulation of Pedestrian Kinematics before Impact with a Vehicle using an Active Pedestrian Human Body Model

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I. INTRODUCTION

A vehicle impact with vulnerable road users (VRUs) can happen in a large variety of possible body postures or movement situations. The human musculature and respective muscular activity lead to joint stiffness, which determine the resulting kinematics in the crash scenario [1]. Therefore, we suggest to apply an active pedestrian human body model (APHBM) to investigate how different muscle activity states influence the body kinematics and, subsequently, the risk of injury for VRUs. To our knowledge, there is no full-body finite element (FE) APHBM freely available on the market, but preliminary work exists [2]. In this study, biofidelic active muscle elements composed of extended Hill-type material (EHTM) [3] were inserted into the modified THUMSv3 pedestrian model in LS-DYNA. Proposed muscle material provides a more realistic eccentric force-velocity relationship and a better damping characteristic compared to standard *MAT_MUSCLE (*MAT_156) available in LS-DYNA. Additionally, EHTM has an integrated muscle length-based controller. To enable directional control of the model, however, an additional level of joint angle control is required. This mid-level control is necessary to reproduce human movement in the pre-crash phase, i.e. before the impact.

In order to reduce the simulation runtime and to tune the controller parameters, a multi-body (MB) model "twin" with similar body size, weight and muscle parameters was created in the in-house environment demoa [4]. The model has a joint angle controller with additional open-loop control and is capable of reproducing simple gait movements recorded during lab experiments.

As a result, this study focuses on the hybrid approach, to use a simplified MB model to preprocess the movement with the subsequent transfer of the resulting control signals, including muscle parameters, to a FE model. Human kinematics is reproduced up to the vehicle impact. Then a transfer of the body position, loads and muscle activation state will take place to solve the in-crash simulation with full modified THUMSv3. Due to the high complexity of the APHBM muscular system, having 628 muscles, only the results for the lower extremities, with 120 controlled muscles, are discussed in this short communication.

II. METHODS

Experimental Data

Motion capture data were taken from the *Human3.6M* (H36M) dataset [5] and used to set the target values for the angle controller. Data were sampled at a rate of 100Hz and transferred to a controller cluster for a movement duration of 3 s. For the lower extremities, hip flexion, hip abduction and knee flexion were monitored. Hip rotation was specified with a constant value of 0° and the positions of the ankle joints with -5°.

Computational Modelling

MB APHBM was created with the in-house preprocessor calcman[®] [4] and adapted to the anthropometry and initial position of the modified THUMSv3 AM50 pedestrian model [1]. The publications of Arnold [6], Kura [7] and Klein Horsman [8] were used as a reference for the muscle parameters and their routing path on the legs. The muscle controller in the MB APHBM has a separate PID controller for each angle. Other important parameters are the effective lever arm positions of the insertion, deflection and origin points relative to the joint centre. Simulation in the demoa [4] was done for 3 s real-time, whereby the joint angles described under *Experimental Data* were transmitted to the controller at a fixed frequency. The joint angle controller processes the deviation of a given joint angle position (deg) to the currently expected angular position. As a result, a lever arm dependent stimulation signal (u) is transmitted to each muscle interacting over a joint.

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The stimulation signal u is added to a fixed "open loop" stimulation value, which can be used to implement muscle co-contraction phenomena. The open stimulation value was specified in this study as 1% of the maximal value for all muscles to ensure a natural minimum activity. The demoa model has muscle material proposed in [9], which served as a basis for LS-DYNA material implementation [3]. Consequently, the resulting muscle activation can be directly transferred between the two models, assuming that all the other model



Fig. 1. APHBM in (a) demoa and (b) aligned with THUMSv3.

parameters are similar. To ensure the structural consistency of the models, the positions of the muscle insertion, origin and deflection points were transferred directly from demoa to LS-DYNA using the same global coordinate system. Physiological muscle routing was implemented with the fixed via-points method using *PART_AVERAGED keyword. Furthermore, insertion and origin points were bound to the nearest three nodes of the SHELL surface of the skeleton by *CONSTRAINED_NODAL_RIGID_BODY (*CNRB) keyword. Deflection points were bound to the next five nodes of the skeletal structure. Finally, the resulting muscle activity (ACT) calculated by the controller in demoa was transferred to LS-DYNA as *DEFINE_CURVE (*DCID) for each muscle, with a downsampling by a factor of 10, ending up in an approximate resolution of 100 points/second.



Fig. 2. Muscle forces in the lower extremities of the modified THUMSv3 model.

III. INITIAL FINDINGS

Despite the complex muscle structure of the lower extremities, with 60 actively controlled muscles per leg, the angle controller in demoa was successfully able to perform a gait movement simulation. Appropriate ACT signals were transferred to LS-DYNA, resulting in not only comparable kinematic results but also similar joint torques and muscle forces for both models. See Fig. 2 for the predicted muscle forces over the first 2 s in the FE APHBM.

IV. DISCUSSION

Initial results of the study show that the suggested hybrid approach of the movement control transfer through muscular activity signal from a MB to a FE environment worked as expected.

However, currently the method proposed has some limitations. First, the simulation was done without considering ground contact forces, because it is currently uncertain whether the given contact models are sufficient to model natural ground contact with a slip-stick distinction and how this would affect the model kinematics. Secondly, the simulations were done with fixed muscle deflection points. An extension of the LS-Dyna code by a complex deflection algorithm, as shown by [10], would be useful. Thirdly, the present MB model lacks representation of the FE soft tissues properties affecting the joint's stiffness. Further studies with the FE model are required to assess the influence of the soft tissues, as well as the actual effects of muscle-induced joint stiffness. In particular, the "relaxation of the FE model" by varying the soft tissues' material properties, as proposed in [11], can provide a better understanding of varying joint and body stiffness and their influences on pedestrian safety. Finally, it is planned to implement a complete controller code in the LS-DYNA environment in the future, avoiding additional MB software use.

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