Developing and Implementing New Flesh Materials based on Human Tissue Data for GHBMC M50-O

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I. INTRODUCTION

Owing to the focal nature of an impact to the human body, soft tissues can experience extreme mechanics during impact with high deformation rates (up to 1000 1/s), stress wave propagations (around 1500 m/s), and large deformations (with stretches 100% elongation and true shear strains over 50%). The mathematical constitutive models commonly used to represent this behaviour are limited, and their implementation into existing computational codes for numerical analysis (which are commonly developed using principles based on small strain, quasi-static assumptions) often result in unstable or inaccurate outcomes.

Flesh, which includes skin, muscle, and adipose tissues, plays a significant role in how the body responds to external forces. During a blunt impact, the flesh absorbs and distributes forces across the body, influencing how the underlying skeletal structures and organs are loaded. Thus, accurate modeling of human flesh tissue mechanics ensures that the internal loading of human body models (HBMs) is biofidelic. The existing flesh materials of the current GHBMC models are not based on human tissue data, because until recently human flesh tissue data covering a range of strains and strain-rates applicable to impact simulation did not exist. This study utilises new human tissue data collected at UVA to formulate constitutive models of human flesh (skin, adipose tissue, passive muscles). These constitutive models were adapted for the same element types and same level of mesh size currently adopted in GHBMC M50 detailed occupant model. We then implemented the new material data and techniques into modeling flesh in GHBMC model and tested these new materials for stability and accuracy under various impact conditions, including component-level and whole-body tests.

II. METHODS

Multiple modes of dynamic mechanical characterisation data of human skin [1] and adipose tissue [2] were used to determine the parameters for the Ogden hyperelastic constitutive model [3] with the quasi-linear viscoelastic formulation [4] (Table I). While the Ogden hyperelastic model does not represent anisotropy typically present in soft tissues, it is a constitutive model that is natively available in most finite element solvers.

TABLE I
OGDEN HYPERELASTIC MATERIAL PARAMETERS FOR VARIOUS FLESH MATERIALS

Material	μ_1 α_1	μ_2 α_2	μ_3 α_3	G_1 β_1	G_2 β_2	G_3 β_3	G_4 β_4	G_5 β_5
Skin	24.1e-3 2.830	-51.0e-3 2.002	55.55e-3 0.613	0.043 1e-4	0.314 1e-3	0.240 1e-2	0.277 1e-1	0.000
Adipose	1.467e-3 3.36e-2	7.201e-9 16.40	-1.89e-4 0.258	0.045 1e-4	0.585 1e-2	0.290 1e-1	0.290	0.000
Muscle	1.25e-7 8.000	-2.50e-7 -4.000		0.055 1e-4	0.098 1e-3	0.041 1e-2	0.557 1e-1	0.232 1e-0

Note: μ terms have units of GPa, β terms have units of ms^{-1}, and a and G terms are unitless.

These materials were implemented into a model that represented the impact tests performance on human flesh via a 4.5 m/s dynamic indentation test [5]. Through an inverse fitting, parameters for the Ogden model were determined for the muscle material (Table I) using the indentation test data (Fig. 1A) and verified with literature data [6-7]. These constitutive models were implemented into the M50-O v6.1 model for all skin, adipose, and muscle parts, except those in the head and neck, which were developed and calibrated using neck load cases.

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With the new materials, whole-body simulations were performed under impact (4.5 m/s Shoulder Impact [8], 6.0 m/s Abdomen Bar Impact [9], 6.7 m/s Abdomen Oblique Impact [10], 6.7 m/s Frontal Chest Hub Impact [11]), gravity settling [12], and frontal crash with reclined seating [12]. Model results were plotted against experimental corridors, and CORA [13] was calculated for each impact case.

III. INITIAL FINDINGS

The new skin material is stiffer than the skin material in the GHBMC M50-O v6.1 model, but the new bulk materials (adipose and muscle) were softer. The new materials matched the experimental indentation test data (Fig. 1A) and were very robust under the extreme levels of deformation associated with that test. Overall, the new materials caused a slightly stiffer whole-body response in impact (Fig. 1B and C) but a softer quasi-static response in gravity settling. CORA values did not change substantially, indicating that the change in flesh material did not significantly alter the existing model's biofidelity. In all impact scenarios, the model was stable.

Preliminary analysis of the reclined seating simulation demonstrated a substantial improvement in whole-body kinematics with the updated materials. The excessive forward flexion of the upper torso of the GHBMC model in this test condition, as reported in a previous study [14], was reduced with the new materials, resulting in the updated model having a more comparable response to the experimental PMHS. This was likely due to the increased stiffness of the nonlinear portion of the skin material, but further investigation is needed to quantify the improved biofidelity of the updated model.

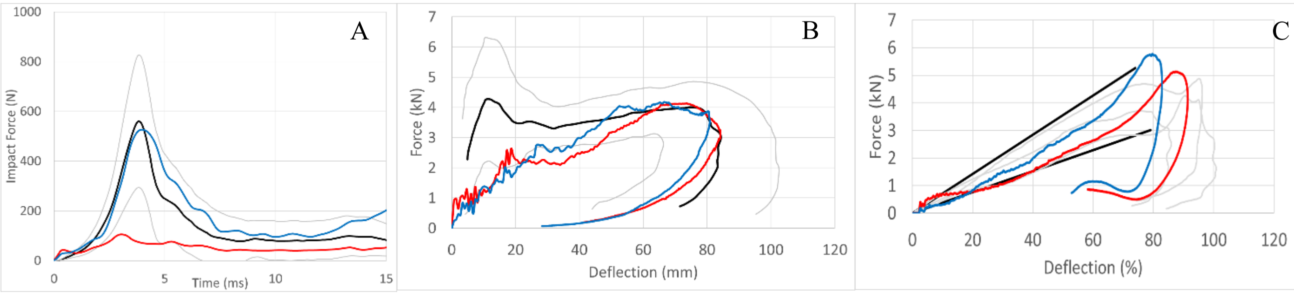

Fig. 1. (A) 4.5 m/s human flesh indentation response. (B) 6.7 m/s PMHS frontal chest hub impact. (C) 6.0 m/s PMHS abdomen bar impact. Response with new materials in blue, v6.1 materials in red, and experimental response in black/grey.

IV. DISCUSSION

By developing new skin, adipose, and muscle tissue constitutive models derived from human tissue experiments, and by incorporating these models into the GHBMC M50-O model, we have improved the biofidelity of the M50-O without drastic changes to overall biomechanical response of the model and its previous validation. Preliminary simulation results also indicate a substantial improvement in the occupant kinematics in a frontal crash with a reclined seating position with the new material models. Incorporating this biomechanical data will allow researchers and engineers to perform more realistic impact simulation of localised impact response to facilitate the design and evaluation of protective gear and safety systems.

V. REFERENCES

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