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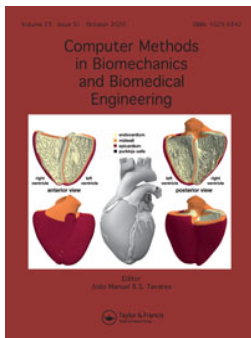
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# Development of a finite-element muscle model accounting for transverse loading

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## 1. Introduction

To our knowledge, FE neck models including active 1D or 3D muscles have been mainly used for impact simulations and have yet to be used to investigate the effects of volumetric deformation, transverse loading and stiffening interactions between muscles, which may affect posture and task performance in the daily life. Muscle efficiency has been experimentally assessed under the influence of transverse loading in experiments on the rat (Siebert et al. 2014) and human (Siebert et al. 2018) gastrocnemius muscles. In these studies, the authors reported changes in transverse expansion of the muscle and a reduction of the axial contraction force with increased external transverse loading, which would seem to work adversely to the structural transverse stiffening and stabilization effects between muscles or muscles and bones that are generally associated with contraction and bulging. In order to improve the bio-fidelity of a FE neck model (Howley et al. 2014) we investigated the implementation of a continuum passive/active law to account for such transverse effects in a single FE muscle model.

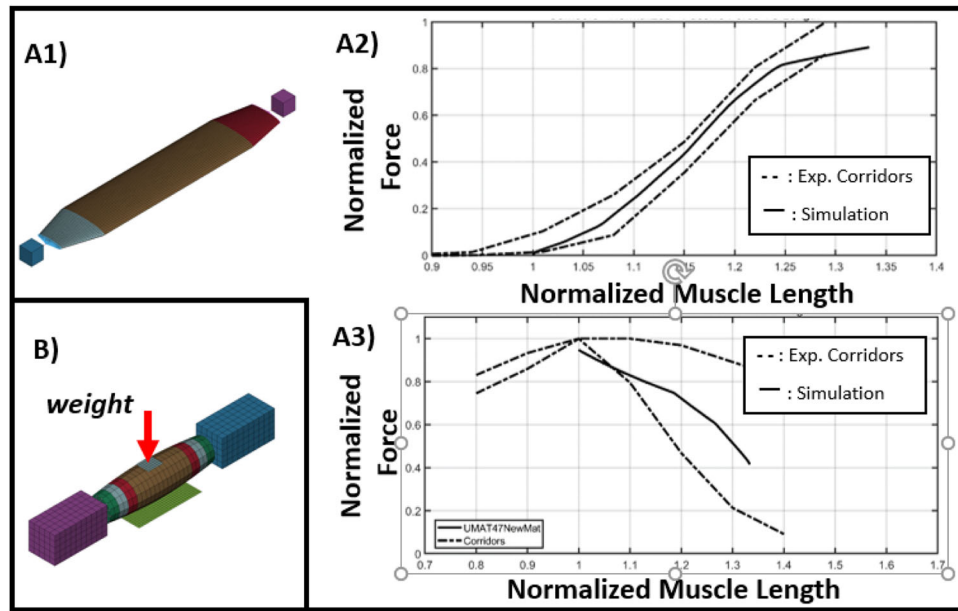
## 2. Methods

A 3D continuum hyperelastic transverse isotropic law, commonly used in the literature (Blemker et al. 2005; Röhrle et al. 2017), was implemented in the FE code LS-DYNA<sup>®</sup> (LS-DYNA 2018) as a user defined material routine, where the stress combines the contribution of a Mooney–Rivlin function and of a unidirectional Hill-type model. A calibration of the axial active-passive force-length behaviour was first performed (case A) by modelling the simplified geometry and material properties (Figure 1(A1)) of a rabbit

*Tibialis Anterior* (TA) muscle and by simulating the experimental behaviour measured during a sequence of passive lengthening, isometric and eccentric active contractions reported by Davis et al. (2003) (Figure 1(A2, A3)). A cylindrical fusiform muscle (Figure 1(B)) with generic human neck muscle material properties was then modelled to reproduce the experimental protocol from (Siebert et al. 2014) where an isometric contraction was applied under transversal weight-bearing loading conditions (Figure 1(B)/case B). In this case the activation level was controlled to try and reproduce the phenomenological transverse/longitudinal force coupling that had been observed by Siebert et al. and to investigate its quantitative effects on a representative neck muscle. The possible mechanistic influence of the presence of an aponeurosis on this coupling was also investigated by extruding and attaching tension-only shell elements to the outside mesh of the model. Young's modulus of the aponeurosis where chosen ranging from 0.1 MPa to 1 GPa (a wider range of conjunctive tissue stiffnesses than reported in the literature), to assess the model's sensitivity to this parameter. The axial force generated by the muscle, the lifting height of the weight (measured as its relative displacement induced during the contraction-only phase) were output and compared to the experimental results (Table 1).

## 3. Results and discussion

Preliminary results for case A simulations compared well with the experiments, the passive lengthening generating a typical exponential axial force (Figure 1(A2)) and the active component producing a bell shape force-length curve (Figure 1(A3)). This calibration against experimental data (Davis et al. 2003) is still in progress. Regarding case B: the difference in muscle height ( $\Delta H$ ) decreased, while the muscle width increased during a sequence of passive weight application followed by an active contraction. However, neither the lifting height ( $\Delta h$ ) nor the axial force produced by the muscle during the active contraction phase showed a significant decrease with the weight, as reported in the experiments. Acknowledging for both possible experimental reporting and modelling limitations discussed below: this means that the model was unable to reproduce by



**Figure 1.** (A1) Rabbit TA model; (A2 and A3) active and passive component force-length results; (B) simulation of Siebert et al. (2014).

**Table 1.** Reported (Siebert et al. 2014) and simulated lifting heights ( $\Delta h_{\text{corr}}$ ) and vertical muscle height change ( $\Delta H_{\text{corr}}$ ) for the same axial force decrease ( $\Delta F$ ) and different plunger weights.

Experiments			Simulations		
$\Delta F_{\text{axial}}$ (%)	Weight (g)	$\Delta h$ (mm)	Weight (g)	$\Delta h_{\text{corr}}$ (mm)	$\Delta H_{\text{corr}}$ (mm)
−4.8	65	1.7	200	3.45	3.13
−7	115	1.3	400	3.27	2.73
−8.9	165	1.1	800	3.16	2.20
−10.5	215	0.8	1600	2.84	1.28
−12.7	265	0.6	3200	2.19	−0.24

mechanical means only the experimental loss in efficiency of the longitudinal force production exhibited by Siebert et al. in the presence of a transversal load. A means to achieve exactly the same relative decrease  $\Delta F_{\text{axial}}$  in the axial force was found by artificially reducing the maximal (MVC) force of the FE muscle model based on the real-time measurement of the lateral load. Doing this correction also resulted in being able to reproduce the decrease in lifting height reported in the experimental data (Siebert et al. 2014). Results ( $\Delta h_{\text{corr}}$ ,  $\Delta H_{\text{corr}}$ ) presented in Table 1 account for this correction.

Regarding the modelling of the aponeurosis: its presence did not fundamentally modify the above described behaviour. However, including it (and increasing its stiffness) resulted in a decrease in both axial muscle force production and lateral bulging.

Using a commonly described active–passive 3D continuum material law, our FE muscle model showed an adequate behaviour in axial loading. However, in order

to reproduce the experimental transverse/longitudinal coupling behaviour previously reported, it required some additional control of the activation. Limitations of our single muscle model include that it does not account so far for the significant pennation angle of the gastrocnemius muscle ( $30^\circ$ ). Further, as muscle fibres' density is not currently controlled for in the material model, transverse geometrical effects resulting in e.g. a dynamic decrease or increase in the PCSA induce slight but uncontrolled changes in the axial muscle force. Further investigation and refinement of the model is thus also required regarding e.g. the modelling of the transverse behaviour of the 3D passive component, the modelling of the aponeurosis and of its attachment to the perimysium.

Finally, the initial positioning and initial shape of the muscle following stabilization under its own weight could not be precisely informed from the experimental publication in the model. Given the small levels of measured experimental lifting heights, we expect that these initial conditions may have had a significant influence in the compared assessment of both experimental and numerical results.

#### 4. Conclusions

Notwithstanding some necessary improvements, it is nevertheless expected that the quasi-incompressible behaviour that is mainly responsible for the model's expansion during active contraction will not be able to fully model the decreased muscle efficiency (axial force) that appears to be induced by an external

transversal load (Siebert et al. 2014). This complex two-way coupling may be influenced by unknown physiological phenomena, which we were only able to account for by artificially controlling the activation level based on the transverse loading. In the prospect of better accounting for macroscopic contact stiffening effects in such models, there is much need for further experimental investigation on these macroscopic coupling effects and on the assessment of transversal muscle force generation.

### Disclosure statement

No potential conflict of interest was reported by the authors.

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**KEYWORDS** muscle; FE modelling; transverse loading

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