A novel active lumbar spine muscle model

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Introduction

Mechanical factors have a significant influence on the degenerative progression of spine tissues. Among the different approximations proposed, numerical models are the particularly suitable to assess the specific contribution of these factors at the tissue or organ scale, but require the prediction of reliable boundary loads generated at the system scale, which remains unsolved. Since the spine musculature transfers most of the kinematical forces to the vertebrae and intervertebral discs, the careful constitutive modelling of muscle activity would stand for a significant improvement of spine models. Yet, no constitutive models have been reported for lumbar spine to our knowledge, despite the presence of particularly high mechanical loads in this area. As such, this study proposes to address this contribution by developing a novel active lumbar spine muscle model and assess its potential to be calibrated specifically to simulate the action of the lumbar spine muscles.

Materials and Methods

A modified quasi-incompressible transversely isotropic hyperelastic constitutive formulation was adopted for the passive and active behavior of the lumbar muscles. The model was based on a modified hyperelastic model originally proposed for cardiac tissues [1] and a similar formulation used previously for the biceps brachii muscle [2,3]. The embedded muscular fibres accounted for both a passive (Eq. 1) and an active (Eq. 2) fibre stress term, where $\bar{\lambda}_f$ is the uncoupled fibre stretch ratio and T_0 is the muscle peak stress. The active part depended also on a strain-like quantity ζ^{CE} that was a function of both optimal and measured rest fascicle length and controlled muscle activation through deformation thresholds (Eq. 3). The whole constitutive relation was the additive contribution of the fibre and of the matrix deviatoric and volumetric stresses (Eq. 4), where *J* is the Jacobian determinant, *K* is the matrix bulk modulus, *G* is the matrix shear modulus, *N* is the undeformed fibre direction and *C* is the right Cauchy-Green strain tensor.

$$f_{PE}(\bar{\lambda}_f) = T_0 \begin{cases} 4(\bar{\lambda}_f - 1)^2, & \bar{\lambda}_f > 1 \\ 0, & otherwise \end{cases} \tag{1}$$

$$f_{SE}(\bar{\lambda}_f, \zeta^{CE}) = T_0 \begin{cases} 0.1\{exp[100(\bar{\lambda}_f - 1 - \zeta^{CE})] - 1\}, & \bar{\lambda}_f > 1 + \zeta^{CE} \\ 0, & otherwise \end{cases} \tag{2}$$

where $$\zeta^{CE} = \frac{L_0^{CE}}{L_0^M}\left(\frac{L^{CE} - L_0^{CE}}{L_0^{CE}}\right) = \frac{L^S}{L_0^S}\left(\frac{L^{CE} - L_0^{CE}}{L_0^{CE}}\right) = C_{CE}\left(\frac{L^{CE} - L_0^{CE}}{L_0^{CE}}\right) \tag{3}$$

$$S = K\ln(J)C^{-1} + \left[f_{PE}(\bar{\lambda}_f) + f_{SE}(\bar{\lambda}_f, \zeta^{CE})\right] \cdot \left[J^{-\frac{2}{3}}\bar{\lambda}_f^{\ -1}(N \otimes N) - \frac{1}{3}\bar{\lambda}_f C^{-1}\right] + \frac{G}{2} \cdot \left(2J^{-\frac{2}{3}}I - \frac{2}{3}\bar{I}_1 C^{-1}\right) \tag{4}$$

A single unidirectional element (*L*=40 mm, *A*=100 mm^2) was used to assess the implemented model by means of user-defined material subroutines. 30% traction and 20% compression strains were imposed. Material constants matrix for the quasi-incompressible behaviour and fibre passive resistance were derived [1,2], while muscle peak stress was based on [4]. For the active fibre stress, morphometry-based estimations for the erector spinae (ES) (Longissimus thoracis pars lumborum and Longissimus thoracis pars thoracis, LT, Iliocostalis lumborum pars lumborum, IL), the multifidus

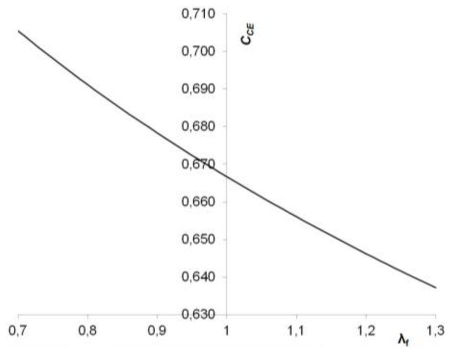
(MF) and the psoas major (PS) muscles based on [5] were used to calculate the parameter C_{CE} (Eq. 3) and to simulate muscle activation (Table 1). The fascicle stresses resulting from the use of these initial parameter values were analysed and the influence of C_{CE} on muscle stress predictions was further parametrically explored in order to assess the possibilities of model calibration based on muscle activation criteria. Additionally, the relative contribution of the different terms of the constitutive model was quantified.

Table 1. Morphometry-based values for C_{CE} for lumbar muscles.

	MF	LT	IL	PS
C_{CE}	0.811	0.825	0.846	1.111

Results and Discussion

Stress analysis of the model indicated that the initial C_{CE} values based on the literature (Table 1) were not able to induce any active response of the muscle due to a systematic overestimation of ζ^{CE} at any fibre stretch ratio. Further parametric analysis showed that C_{CE} values below 0.637 in traction and above 0.706 in compression were necessary to predict non-zero active stresses at some point (Fig.1a). Thus, alternative values were explored by keeping the relative activations of the muscles modelled as suggested by the relation of ζ_{CE} to the functional geometrical characteristics of these muscles: 0.465/0.706, 0.473/0.718, 0.485/0.737 and 0.637/0.967 (traction/compression) for MF, LT, IL and PS, respectively.



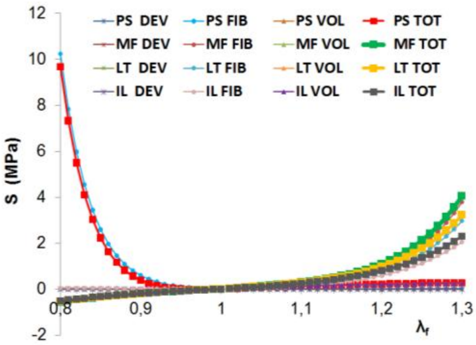


Fig.1. (a) Range of parameter C_{CE} for active stress calculation, (b) 2nd Piola-Kirchhoff stress for MF, LT, IL &PS

Though no data could be found for an exhaustive validation of the model, the new stress results obtained (Fig. 1b) were, for all muscles tested, within the plausible range of values measured for lumbar spine muscle fascicles under λ_f values between 0.9 and 1.1 [6]. PS showed always an antagonistic active behaviour, helping the movement, in contrast to the dorsal muscles that were resisting (Fig. 1b). Under traction, the relative contribution of the different terms in the total stress indicated the dominant role of the active fibre term followed by the passive volumetric term for MF, LT and IL muscles (Fig. 2a). PS volumetric stress response, however, was prominent (Fig. 2a), suggesting the need to assess its dependence upon muscle fibre activation. In the anatomical context of the lumbar spine (Fig. 2b), the compression - traction results suggest that PS would help the flexion - extension of the trunk, respectively, while the other fascicles would resist when activated simultaneously in a lumbar spine muscle model. Thus, the antagonism predicted for the isolated fascicle models is qualitatively realistic when considering the three-dimensional distribution of the lumbar muscle fascicles. Physiological calibration of the model proposed is difficult due to the location of some fascicles, e.g. PS, that usually involves invasive techniques. Nonetheless, explorations through multi-channel surface EMG techniques suggest that such difficulty could be overcome [7].

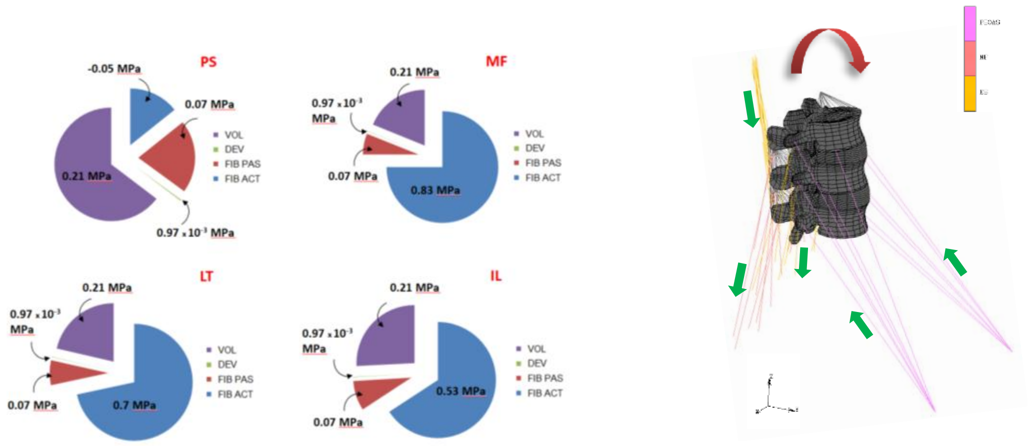


Fig. 2. (a) Relative contributions of stress terms for λ_f=1.2, (b) Three-dimensional L3-L5 lumbar spine model including PS, MF and ES muscles

Conclusion

A new constitutive model for the lumbar spine muscles was proposed and showed to give reasonable predictions for the activation of individual lumbar fascicle models. The limited number of parameters (5) needed in this model is advantageous compared to the corresponding number of 13 parameters involved in other models [8]. The stress predictions of the muscle active response were very sensitive to variation of the parameter C_{CE}. Thus, explorations of the strain range of activation and phenomenological exploration of the parameter ζ_{CE} for the lumbar muscles could provide realistic calibrations for further implementation of the equations in a lumbar spine model. Indeed, these explorations could be performed clinically, e.g. through EMG and/or MRI techniques, opening the way to patient specific calibrations.

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