Using EMG to solve the muscle redundancy problem in musculoskeletal models

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1. Introduction

Computing muscle forces *via* a musculoskeletal model with a purely mathematical approach such as optimisation usually copes with the problem of muscle redundancy, i.e. infinity of solutions for a given configuration. One way to solve this problem is to use cost functions and optimisation methods which reflect actual muscle strategies the best. It often requires an experimental validation realised afterwards. Alternatively, experimental input such as electromyography (EMG), measured during an upfront experiment, can be used to feed the model and to guide the solving towards muscle forces physiologically more correct. For the trunk, if this approach is not new (Granata and Marras 1995), it is still underused (Gould et al. 2021). Therefore, in this study, EMG was used as input of a model of the trunk to compute muscle forces and resulting intervertebral efforts – forces and torques – in the lumbar spine for a static configuration in the Sorensen test posture. To illustrate the necessity of EMG as input, a single participant was asked to produce three muscle strategies for the same position. Whereas the mathematical optimisation cannot see any difference in muscle forces, it is expected that the EMG-based solutions reflect the strategies better.

2. Methods

2.1. Experiment

One 28-year-old male subject (175 cm, 90 kg) participated voluntarily in the experiment. He produced three different muscle strategies in the Sorensen test posture. Lying on a table in a prone position with the trunk unsupported, he had to maintain the trunk aligned with the lower limbs while being (i) the most relaxed as possible (relaxed configuration), (ii) the most stressed as possible (stressed configuration) and (iii) in an intermediate level between relaxed and stressed (mid configuration).

EMG signals of back – lumbar (LP) and thoracic paravertebral (TP) muscles, quadratus lumborum (QL), latissimus dorsi (LD) – and abdominal – rectus abdominis, externus obliquus – muscles were recorded during the experiment. Each configuration was performed twice: unloaded and loaded with an external mass equal to 20% of the body mass. Maximal voluntary contractions (MVC) were recorded prior to the exercises for EMG normalisation.

2.2. Model

A multibody model representing the experiment was developed but it included only muscles actually activated during the experiment: LP, TP, QL and LD.

Muscle forces were quantified *via* two methods: an EMG-based deterministic distribution of a global equivalent force between all muscles (Hinnekens et al. 2021) and one based on optimisation. For the latter, a trust-region algorithm for constrained optimisation was used to implement two cost functions: minimising the sum of cubed muscle stresses (s^3) and minimising the largest relative muscle force (mM) $s^3: \ min \sum_i \left(\frac{F_i}{PCSA_i}\right)^3$ & $mM: \ minmax\left(\frac{F_i}{F_{max,i}}\right)$ with F_i the force in the ith muscle fascicle, $PCSA_i$ its physiological cross-sectional area and $F_{max,i}$ its maximum isometric force.

We made the hypothesis that the spine and its surrounding intrinsic muscles were assimilated to a beam whose intervertebral efforts – compression and shear forces and bending torque – are computed according to the beam theory through multibody constrained forces and torques. Intervertebral efforts were consequently overestimated with respect to the reality. Finally, they were translated into equivalent stresses considering an intervertebral disc with a cross-section area of 18 cm^2. This was done for both computation methods.

3. Results and discussion

EMG amplitudes of back muscles recorded during the experiment (Figure 1) showed varied muscle strategies as expected. Loaded cases produced obviously highest EMG amplitudes. For the relaxed and mid configurations, TP were the most activated for both unloaded and loaded cases. For the stressed configuration, LP were the most activated for the unloaded case.

LD was not expected to be as active, particularly for the loaded cases. LD produced the highest EMG amplitude among all EMG amplitudes for the loaded stressed configuration. This high activation of LD in

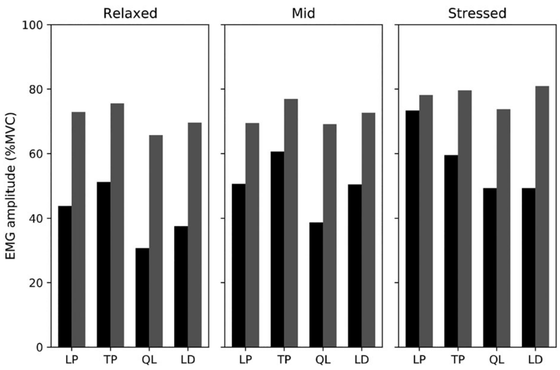

Figure 1. EMG amplitudes (%MVC) of LP, TP, QL and LD for the three configurations and for both unloaded (in black) and loaded (in grey) cases.

this study may come from the bilateral role of this muscle in helping for hyperextension of the trunk (Kendall et al. 2005).

The varied muscle strategies recorded led to varied distributions of muscle forces for the EMG-based approach whereas the optimisation-based approach could not see any differences between the three configurations (not shown).

Figure 2 showed the equivalent stresses associated to compressive and shear forces and bending torque for the four upper lumbar discs computed with the EMG-based approach (relaxed, mid and stressed) and by the optimisation (s^3 and mM). First, the optimisation could not distinguish between the three configurations while the EMG-based approach could. Second, the optimisation produced relatively intense stresses associated to the bending torque at L3/L2 and L2/L1 levels, particularly for the loaded case but also produced lower or relatively similar stresses for other lumbar levels and efforts.

The greatest stresses were associated to the bending torque. Positive values indicate intervertebral discs in which posterior internal fibres are in compression while anterior internal fibres are in traction; negative values indicate the opposite. At L3/L2 level, there were smaller stresses for loaded cases (relaxed, mid and stressed) than for unloaded cases probably due to the zero-point crossing of the bending torque around L3/L2 level for the loaded cases.

For the unloaded case (non-hatched bars in Figure 2), the mid configuration produced the greatest stresses in shear force and bending torque. It might be due to the EMG-based approach which evenly distributed muscle forces between LP and LD

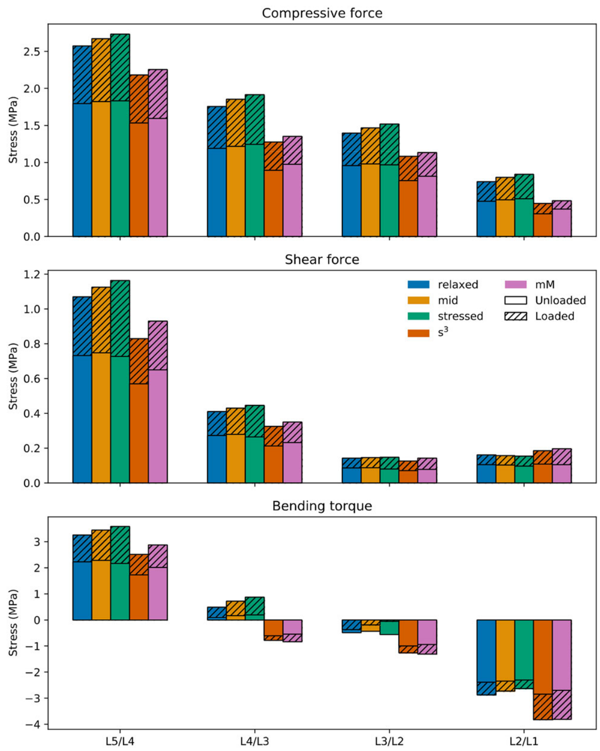

Figure 2. Stresses in the four superior lumbar discs corresponding to (top) compressive and (middle) shear forces and (bottom) bending torque. Non-hatched and hatched bars represented unloaded and loaded cases respectively.

(see Figure 1) while for the stressed configuration, LP were favoured. Given the geometry of LD fascicles, it resulted in greater stresses in shear force and bending torque.

4. Conclusions

This study aimed at highlighting how relevant EMG can be as input for musculoskeletal models, especially when solving the muscle redundancy problem for varied muscle strategies achieved by the same subject.

The EMG-based approach had some limitations, showed by highest stresses of the mid configuration for the unloaded case. Besides, the deterministic muscle force distribution was based on a non-negative linear regression computed on two masses only, resulting in a small increase for LP when adding the load.

Future work will focus on embedding EMG input in the optimisation process itself (Granata and Marras 1995) to make it more physiologically relevant, by modulating its cost functions and/or constraints so as to discern clearly different muscle strategies.

References

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