LUMBAR SPINAL MUSCLE ACTIVATION SYNERGIES PREDICTED BY MULTI-CRITERIA COST FUNCTION

INTRODUCTION
Because there are many more muscles which cross the lumbar spine (>100) than there are degrees of freedom of the lumbar spine (36) there is a highly redundant number of strategies that could be used to control the muscles. It has been assumed that the central nervous system (CNS) uses a physiological cost function of the spinal or muscle forces, or spinal displacements in establishing the appropriate muscle activation strategy, compatible with equilibrium and other physiological constraints.

We analyzed lumbar spinal muscle synergies by comparing EMG data from human subjects with an analytical model whose cost function is the weighted sum of five physiological criteria: (1) minimize global spine displacements; (2) minimize intervertebral motion; (3) minimize intervertebral forces; (4) minimize the sum of cubed muscle stresses (which correlates with muscle fatigue) and (5) maintain stability (i.e. prevent spinal buckling).

METHODS
Fourteen subjects (8 male, 6 female; ages 20-26 years; body mass 75±14 kg) were studied with IRB approval and informed consent. They performed ramped efforts up to a maximum effort isometrically against resistance provided by a harness around the thorax connected to a cable with a load cell. The angle of pull was varied in 45° increments from 0° to 180° to the anterior direction. Their maximum efforts (extension) averaged 701 N. EMG electrodes recorded signals from six right and left pairs of muscles (rectus abdominis, internal and external obliques, longissimus, iliocostalis and multifidus). For multifidus, fine wire electrodes were inserted intra-muscularly. The other electrodes were surface electrodes with integral high input-impedance preamplifiers. The change in muscle activation with effort for each muscle at each angle was expressed as the slope of the RMS EMG-force relationship obtained by linear regression analysis, since these relationships were found to be essentially linear during the ramped voluntary increase of load.

The analytical model was specified in terms of muscle geometry (90 symmetric muscle pairs), spinal shape and the stiffness properties of motion segments [1,2]. The weight of the trunk was considered to be 340 N acting 60 mm anterior to the T12 vertebral body center. The external loading was specified as a horizontal force from zero to 300 N in steps of 50 N acting at angles of 0, 45, 90, 135 or 180° to the anterior direction, in order to simulate experiments with human subjects. In the sub-components of the cost function, displacements and rotations were weighted such that 1 mm of displacement was weighted the same as 1° of rotation. The relative weights of the muscle stress and stability components of the objective function were weighted 1000 times more than the displacement components, to take into account their differing units of measure. Simulations were performed using the MINOS optimization program (Operation Research Laboratory, Stanford University, Stanford, CA) with the muscle forces as variables. The muscle forces were bounded between 0 and 460 kPa multiplied by the muscle physiological cross sectional area (PCSA).

Muscle activation slopes were compared between the experimental (EMG) and analytical values after normalizing them by dividing by the maximum muscle force value for the model, and by the maximum recorded EMG value for the experiments. Resulting slopes were in percent-of-maximum-activation/kN. Because the model included 180 muscles (90 pairs), the activity levels were weighted by PCSA for averaging within each anatomical group for comparisons with the experimental results.

RESULTS
The muscle activation slopes determined experimentally (average of 14 subjects and 3 repeat trials) are shown in Figures 1 and 3. All the muscles except multifidus have positive slopes indicating that these muscles, even at angles where they are antagonistic, increased their level of activation with effort.

The muscle activation slope from the model are shown in Figures 2 and 4 in the same format as used for the experimental results (Figures 1 and 3). The agreement was close, except for certain muscles (especially the internal obliques) where the degree of coactivation was less according to the model. Muscle activation patterns included coactivation of antagonist muscles as evidenced by the mostly positive slopes in Figures 1 and 3.

DISCUSSION
As expected, the model with a 5-component objective function developed here gave better agreement with experimental data than models which consider equilibrium at only one intervertebral level, and which only consider one physiological objective when estimating the muscle activation strategy. It is not feasible that the CNS would attempt to optimize a single physiological factor to the exclusion of others. The linearity of EMG-force relationships suggests that the activation strategy of the trunk muscles is generally preset, and continues unchanged during a graded increase in effort. The idea that there might be a ‘common drive’ of the trunk muscles giving a constant set of synergies with changing effort is attractive. It would reduce the number of control channels required to coordinate the large number of trunk muscles. The realism of this analysis is improved by inclusion of stability and other physiologically necessary components in the optimization cost function for calculating muscle activation patterns.

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REFERENCES