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VALIDATION OF THE OPENSIM LIFTING FULL-BODY MODEL TO EVALUATE SPINAL LOADS DURING LIFTING TASKS AND OTHER ACTIVITIES OF DAILY LIVING

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INTRODUCTION

Low back pain is a common and costly health condition that is related to occupation and activities of daily living [1, 2]. Tasks that involve forward bending and lifting have been identified as risk factors for the development of low back pain, due to the resulting large spinal loads [3]. Direct measurement of *in vivo* spinal loads is challenging, as it requires invasive measurements of intradiscal pressure (IDP) [4] or invasive surgery to implant an instrumented vertebral body replacement (VBR) in patients affected by spinal disorders [5]. One alternative to direct measurement of *in vivo* loads is computational modelling. Musculoskeletal models can be used to simulate the complexities of the spine and evaluate spinal loads during lifting or similar tasks [6,7]. However, most musculoskeletal models are not open source (i.e. unavailable to the biomechanics community), and there is currently no validated full-body model for the evaluation of spinal loads during lifting on the OpenSim platform (SimTK, Stanford, CA) [8]. Therefore, the aims of this study were to (1) adapt the trunk musculature and movements of an existing OpenSim jogging full-body model to lifting analysis requirements, and (2) to validate the resulting model for the evaluation of lumbar spine loading during symmetrical and asymmetrical lifting tasks.

METHODS

The lifting full-body (LFB) model was developed by modifying the joints and muscle properties of the Full-Body Lumbar Spine (FBLS) model [9] to suit the analysis of lifting tasks. These modifications to the model were validated using sensitivity analysis and comparisons with reported experimental data.

Three healthy male participants performed three repetitions of four lifting tasks in several loading conditions: two-handed stoop (2ST), two-handed squat (2SQ), one-handed stoop (1ST), and braced-arm-to-thigh (BATT). Participants also performed static tasks consisting of unloaded and loaded (5 kg mass in each hand) standing with straight legs and holding trunk flexion angles of 0°, 10°, 20°, and 30°. Kinematic and kinetic data were collected with a 12-camera Vicon motion analysis system (100Hz, Oxford Metric, UK) and two force platforms (2kHz, AMTI, USA). A three-axis load cell (2kHz, Kistler, SUI) secured to the thigh directly above the knee measured the bracing forces applied by the hand.

Model outputs were verified using direct comparison between estimated muscle activations and experimentally measured electromyography (EMG) from the participants in this study, and indirect comparison with reported *in vivo* IDP and VBR measurements on different participants. Simulations were performed in OpenSim (version 3.3); the force(s) created by the box held in the hand(s) was modelled by incorporating the mass properties of the box to those of the hand(s).

EMG muscle activation comparison was evaluated using a cross-correlation analysis, while comparisons with IDP were evaluated by a correlation analysis. Qualitative comparisons with VBR measurements were performed.

RESULTS AND DISCUSSION

The resulting LFB model comprised 30 segments, 29 degrees-of-freedom (six lumbar joints with three degrees-of-freedom each), and 238 Hill-type musculotendon actuators (trunk musculature only) (Fig 1). All maximum isometric muscle forces were increased, with a maximum muscle stress (MMS) of 100 MPa, to ensure the model could satisfy equilibrium for all lifting activities evaluated in this study. On evaluation of the model, the corresponding maximum isometric extensor moments for the model remained within the reported ranges [10].

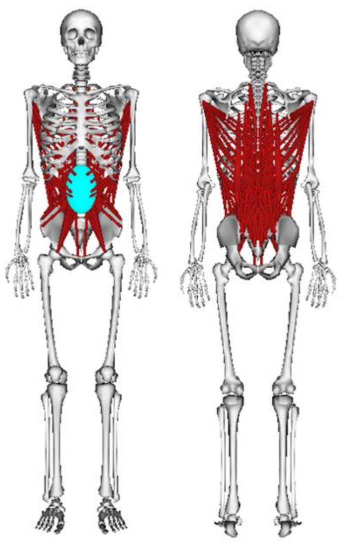

Fig 1: The LFB model with 238 musculotendon actuators for the trunk musculature.

The LFB model predictions for muscle activation agreed well with the recorded experimental EMG signals for the lumbar and thoracic erector spinae, both in the timing and pattern of the signals (Fig 2). The peak cross-correlation values for the back muscles were all r>0.82, and reached values as high as r=0.93.

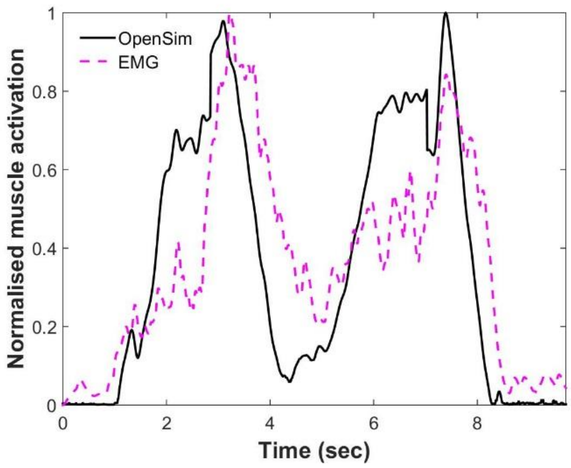

Fig 2: Measured EMG and muscle activation estimated by the model for the right lumbar erector spinae during a squat.

The estimated IDP values agreed strongly with the *in vivo* measurements (R^2=0.87). In addition, vertebral loading estimates increased as the mass lifted and forward bending also increased during dynamic lifting tasks, indicating the model can appropriately estimate changes in spinal loading during loaded and unloaded dynamic trunk movements.

The agreement between the normalised VBR measurements and the model estimates was low, but the same trends were observed: increasing intervertebral forces with forward bending and higher intervertebral forces when the load was held in hands compared to when participants were empty-handed. However, several limitations are associated VBR implants and the absolute values of the measurements must be evaluated with care.

CONCLUSIONS

The LFB model is an appropriate tool to non-invasively evaluate changes in lumbar loading during symmetrical and asymmetrical lifting tasks and other ADLs that involve frequent forward bending. The LFB model is available on the OpenSim repository.

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