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A novel stability-based EMG-assisted optimization method for the spine

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ABSTRACT

Traditional electromyography-assisted optimization (TEMG) models are commonly employed to compute trunk muscle forces and spinal loads for the design of clinical/treatment and ergonomics/prevention programs. These models calculate muscle forces solely based on moment equilibrium requirements at spinal joints. Due to simplifications/assumptions in the measurement/processing of surface EMG activities and in the presumed muscle EMG-force relationship, these models fail to satisfy stability requirements. Hence, the present study aimed to develop a novel stability-based EMG-assisted optimization (SEMG) method applied to a musculoskeletal spine model in which trunk muscle forces were estimated by enforcing equilibrium conditions constrained to stability requirements. That is, second-order partial derivatives of the potential energy of the musculoskeletal model with respect to its generalized coordinates were enforced to be positive semi-definite. Fifteen static tasks in upright and flexed postures with and without a hand load at different heights were simulated. The SEMG model predicted different muscle recruitments/forces (generally larger global and local muscle forces) and spinal loads (slightly larger) compared to the TEMG model. Such task-specific differences were dependant on the assumed magnitude of the muscle stiffness coefficient in the SEMG model. The SEMG model-predicted and measured L4-L5 intradiscal pressures were in satisfactory agreement during simulated activities.

1. Introduction

In the absence of accurate non-invasive *in vivo* techniques, musculoskeletal models are widely employed to compute biomechanical loads on different body joints during static and dynamic activities [1,2]. Three approaches have been proposed to resolve joint kinetics redundancy. First, a pure optimization technique in which muscle forces are estimated by optimizing an assumed cost function such as the sum of squared or cubed muscle stresses [3]. Second, a pure electromyography (EMG)-assisted approach in which muscle forces are estimated from an assumed muscle EMG-force relationship during the task under consideration [4–6]. As the EMG-estimated muscle forces do not necessarily satisfy the moment equilibrium requirements, a subject-specific, common (single) correction gain to all muscle forces is calculated with the aim of minimizing differences between the EMG-predicted and *in vivo*-measured (inverse dynamics) moments over the duration of each task. Third, a hybrid EMG-assisted optimization approach in which the muscle-specific correction gain factors are estimated through an optimization approach thereby perfectly satisfying moment equilibrium requirements [4,7].

However, both EMG-assisted and EMG-assisted optimization approaches make routine assumptions on the surface EMG crosstalk, EMG recording of deep and wide muscles [8], EMG-force relationship [9], processing/normalization of EMG data, and estimation of muscle gains. While the pure EMG-assisted approach has the advantage of considering a common subject-specific gain for all muscles during the tasks under consideration, it fails to perfectly satisfy moment equilibrium requirements, i.e., errors are allowed between the EMG-predicted and *in vivo*-measured moments. On the other hand, the EMG-assisted optimization approach perfectly satisfies moment equilibrium requirements in all anatomical planes during the tasks under consideration, but **it** suggests muscle-specific gains that may vary for a single subject from one task to another. Nevertheless, it has been shown that the EMG-assisted and EMG-assisted optimization approaches provide similar estimates for muscle forces [7] and joint compression [4].

A major shortcoming of the traditional optimization-assisted and both EMG-assisted models is in predicting muscle forces based solely on the equilibrium requirements with no consideration for mechanical stability requirements. While the passive lumbar spine (devoid of muscles) becomes unstable under small compressive

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loads of ~100 N [10], it experiences much larger loads during *in vivo* activities [11] without becoming unstable. That is, trunk muscle activations/co-activations provide stiffness and therefore enhance *in vivo* spinal stability [12,13]. This stabilizing role of muscles cannot be explained by traditional equilibrium-based models. Stability-based optimization-assisted models of the spine have therefore been developed in which trunk muscle forces are estimated while satisfying both equilibrium and stability requirements [14–16]. As an index of stability, the second-order partial derivatives of potential energy of the musculoskeletal model with respect to its generalized coordinates (to produce the Hessian matrix) was constrained to be positive semi-definite, i.e., all eigenvalues of the Hessian matrix remain positive in the optimization algorithm [14–16]. It was shown that muscle activity patterns were more accurately predicted (compared to *in vivo* EMG activities) when stability requirements were considered in the optimization algorithm [15,16].

These stability-based solutions of muscle forces have, however, been implemented only in the optimization-assisted models of the spine [14–16]. This could be due to the premise that EMG-assisted models use physiological signals to predict muscle forces and thereby automatically satisfy stability requirements. However, due to simplifications/assumptions in the measurement of surface EMG activities and in muscle force estimations from EMG data, these models may fail to satisfy stability requirements. Using our EMG-assisted musculoskeletal trunk model [16], we predicted an unstable spine during some static tasks that were in reality stable, i.e., were successfully accomplished by the subject [17]. This could be an indication of an improper prediction of muscle forces by the traditional equilibrium-based EMG-assisted and EMG-assisted optimization models.

The present study aims to develop a novel stability-based EMG-assisted optimization (SEMG) method applied to a musculoskeletal spine model in which trunk muscle forces are estimated by enforcing equilibrium conditions using muscle-specific gain factors [7] constrained to stability requirements in an optimization algorithm. Moreover, the effect of different muscle stiffness coefficient values on predictions of the SEMG model for muscle forces and spinal loads is investigated. Predictions of the SEMG model for muscle forces and spinal loads will be compared with those of the traditional EMG-assisted optimization (TEMG) model in which only equilibrium requirements are met using muscle-specific gain factors. Model predictions for the L4-L5 intradiscal pressure (IDP) are validated against the measured values [11] in the simulated tasks.

2. Materials and methods

2.1. In vivo *experimental study*

EMG data required for estimation of muscle forces and kinematics data required for estimation of posture and external moments were recorded under fifteen isometric tasks [12,18], and then used as input to the musculoskeletal model. Full body kinematics/posture of an asymptomatic male subject (52 years, ~174 cm and ~68 kg) were measured using twelve clusters, each having 4 light-emitting diodes (LEDs) except those on the feet which each had 7 LEDs, located on each foot, thigh, upper arm, forearm as well as one on the pelvis, T12, C7 and head for a total of 54 LEDs (Fig. 1a). Three-dimensional positions of the markers were recorded with a sampling rate of 30 Hz using an Optotrak system (Northern Digital, Waterloo ON, Canada). LEDs were also positioned on the hand load to measure its three-dimensional coordinate. In addition, EMG signals were recorded bilaterally by twelve active surface electrodes (Model DE-2.3, DelSys Inc., Wellesley, MA, USA; bandpass filter: 20–450 Hz; pre-amplification gain: 1000) from back muscles including the longissimus dorsi at

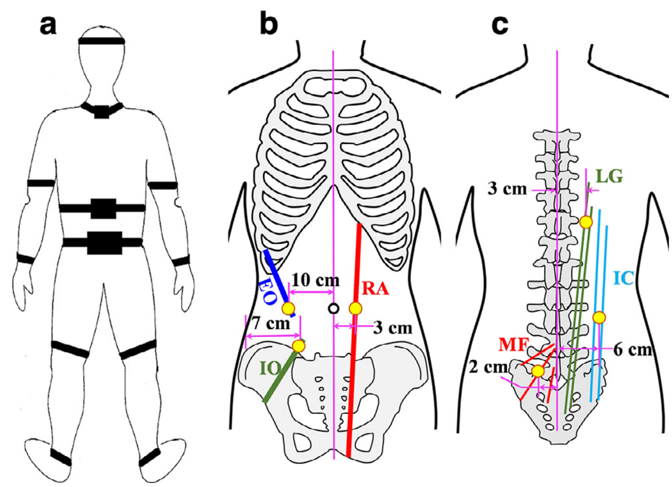

Fig. 1. Schematic representation of the placement of the (a) cluster bands (each having 4 LEDs except the one of foot with 7 LEDs), (b) abdominal EMG electrodes, and (c) back EMG electrodes (solid yellow circle symbols). LG: longissimus dorsi, IC: iliocostalis, MF: multifidus, RA: rectus abdominis, EO: external obliques, and IO: internal obliques.

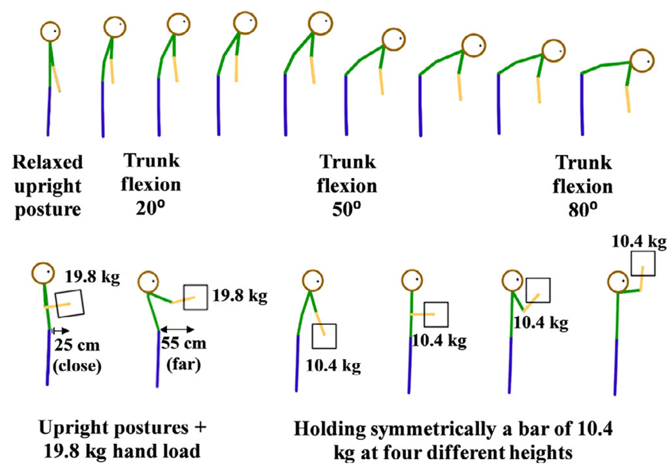

Fig. 2. Schematics of the simulated tasks in upright and flexed postures to estimate muscle forces and spinal loads.

L1 (~3 cm lateral to the midline), iliocostalis at L3 (~6 cm lateral to the midline), multifidus at L5 (~2 cm lateral to the midline) as well as abdominal muscles including the rectus abdominis (~3 cm lateral to the midline above the umbilicus), external obliques (~10 cm lateral to the midline above umbilicus) and internal obliques (~2 cm below and 7 cm medial to the anterior superior iliac spine) (Fig. 1b and c). The raw EMG signals were collected at 1024 Hz and an 8th order bandpass (30–450 Hz) filter was first applied to exclude the electrocardiographic signal and then again to reduce the potential influence of skin movement on the signal. Subsequently, the signal full-wave was rectified and low-pass filtered (single pass, Butterworth) at a cut-off frequency of 3 Hz to get the linear envelope. Artefacts on the linear envelope were detected visually from the signals and removed using cubic splines [12,18]. For normalization, EMG data were collected while the subject stood in an in-house tri-axial dynamometer [12,18] and performed maximum voluntary contraction (MVC) in each anatomical plane (three trials for each exertion). EMG data from each muscle were normalized to its EMG activities during MVC activities.

The subject performed fifteen static tasks (three trials for each task) in upright standing and flexed postures with and without a hand load (Fig. 2). This included: (1) relaxed upright standing