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Trunk musculoskeletal response in maximum voluntary exertions: A combined measurement-modeling investigation

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ABSTRACT

Maximum voluntary exertion (MVE) tasks quantify trunk strength and maximal muscle electromyography (EMG) activities with both clinical and biomechanical implications. The aims here are to evaluate the performance of an existing trunk musculoskeletal model, estimate maximum muscle stresses and spinal forces, and explore likely differences between males and females in maximum voluntary exertions. We, therefore, measured trunk strength and EMG activities of 19 healthy right-handed subjects (9 females and 10 males) in flexion, extension, lateral and axial directions. MVEs for all subjects were then simulated in a subject-specific trunk musculoskeletal model, and estimated muscle activities were compared with EMGs. Analysis of variance was used to compare measured moments and estimated spinal loads at the L5-S1 level between females and males. MVE moments in both sexes were greatest in extension (means of 236 Nm in males and 190 Nm in females) and least in left axial torque (97 Nm in males and 64 Nm in females). Being much greater in lateral and axial MVEs, coupled moments reached ~50% of primary moments in average. Females exerted less moments in all directions reaching significance except in flexion. Muscle activity estimations were strongly correlated with measurements in flexion and extension (Pearson's r = 0.69 and 0.76), but the correlations were very weak in lateral and axial MVEs (Pearson's r = 0.27 and 0.13). Maximum muscle stress was in average 0.80 ± 0.42 MPa but varied among muscles from 0.40 ± 0.22 MPa in rectus abdominis to 0.99 ± 0.29 MPa in external oblique. To estimate maximum muscle stresses and evaluate validity of a musculoskeletal model, MVEs in all directions with all coupled moments should be considered.

1. Introduction

Maximum voluntary exertion (MVE) measurements and musculoskeletal modeling quantify both trunk and muscle strength (Burkhart et al., 2017; Roy et al., 1989; Tsao et al., 2008). The database of MVE moments (for both females and males) can be helpful in risk of injury assessment (Potvin, 2012), functional diagnosis (Dankaerts et al., 2004; Demoulin et al., 2013), performance evaluation/enhancement (Arja et al., 2003) and rehabilitation and treatment evaluations (Stokes, 2011). Previous measurements have recorded highest isometric trunk strength in extension and lowest in axial twist (Azghani et al., 2009; Larivière et al., 2009) and found that various factors such as gender (smaller in females) (Kumar, 1996; Lee and Kuo, 2000; Plamondon et al., 2014), back pain (Dankaerts et al., 2004; Larivière et al., 2003; Ng et al., 2002) and posture (Gravel et al., 1997; Kumar, 1996; O'sullivan et al., 2006) affect measured MVE moments.

There is however no direct method to measure muscle forces and spinal loads; therefore, estimating muscle forces, internal loads and maximum muscle stress (or specific muscle tension) during MVEs is only possible through musculoskeletal (MS) modeling. Earlier studies have used generic (not individualized) EMG-driven (Cholewicki et al., 1995), optimization-driven (Gardner-Morse et al., 1995; Gatton et al., 2011; Jamshidnejad and Arjmand, 2015; Song and Chung, 2004) and kinematics-driven (Arjmand et al., 2008; El Ouaaid et al., 2013) models to investigate internal loads and muscle activities. Subject-specific MS models should however be used to analyze MVE tasks since maximum muscle stress, muscle moment arm and muscle area (all affecting trunk strength) in addition to passive ligamentous properties vary from one person to another. MS modeling along with MVE measurements have been employed to estimate

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maximum muscle stresses at jaw (Pruim et al., 1980), elbow (Buchanan, 1995; Kawakami et al., 1995), wrist (Goislard et al., 2017), ankle (Fukunaga et al., 1996) and trunk (only in extension) (Burkhart et al., 2017) MVEs. However as yet, no study has either investigated trunk responses (internal loads and muscle activities) during MVEs for both females and males or estimated the maximum muscle stress of trunk muscles considering MVEs in all directions.

In the present study, we aim to simulate MVE tasks in a subject-specific model, compare predicted muscle activities with measured EMGs, compute maximum muscle stresses and finally investigate likely differences between females and males in exerted MVE moments and spinal loads. We initially carry out isometric MVE experiments in extension, flexion, lateral and axial directions on 19 asymptomatic young right-handed female and male subjects while recording EMGs of superficial muscles. Furthermore, to explore the accuracy of our geometrically subject-specific nonlinear MS FE model, we simulate MVE tasks, estimate maximum muscle stresses and compare estimated activities of select muscles with measured EMGs.

2. Methods

2.1. Experiments

With approval from our institutional review board and written consent from participants, healthy young right-handed females (9 females; height = 163.4 ± 3.7 cm; weight = 61±4.5 kg; age = 24.1 ± 4.3 years) and males (10 males; height = 174.6 ± 4.2 cm; weight = 72.2 ± 8.7 kg; age = 30.6 ± 6.5 years) performed two trials of flexion, extension, lateral and axial isometric MVEs in a dynamometer at a semi-seated posture (Larivière et al., 2001). During trials (lasting ~8 s), subjects were verbally encouraged to exert their maximal effort while their pelvic and legs were fixed, and their hands were held crossed on the chest. Each trial afterward followed by a two-minute rest. Three triaxial force platforms (Advanced Mechanical Technology Incorporated, model MC6-6-1000, Watertown, MA, USA) collected dynamometer signals at 128 Hz frequency. An EMG acquisition device (model DE-2.3, DelSys Inc., Wellesley, MA) recorded EMG signals of 12 superficial muscles (longissimus, iliocostalis pars thoracic/lumborum, multifidus, external oblique, internal oblique and rectus abdominis) at the frequency of 1024 Hz via surface electrodes placed bilaterally, Fig. 1 (De Foa et al., 1989; McGill, 1991). A band-pass filter (30–450 Hz) reduced the effects of noises and artifacts from EMG signals, and subsequently, root mean squared envelopes of EMG amplitudes were normalized to their recorded maximum root mean squared values during MVE tasks. Data of the trial with the larger primary moment were considered in these and subsequent analyses.

2.2. MS modeling

We simulated MVE tasks in all 6 directions for all 19 subjects in our geometrically subject-specific nonlinear finite element MS trunk model (Ghezelbash et al., 2016a,b). The model includes 126 sagittally-symmetric muscles and computes muscle forces in an optimization- and kinematics-driven framework while taking account of seven individual (T11-T12 to L5-S1) motion segments as shear-deformable beams. Each deformable beam attaches two adjacent rigid vertebrae and represents the stiffness (moment-curvature and force-strain) of an entire motion segment (disc, facets, ligaments and vertebrae). For a given set of prescribed thoracolumbar (T11 to S1) rotations (for details see below), required moments at each vertebral level were initially determined from the nonlinear FE model. An optimization algorithm (with quadratic sum of muscle stresses as the cost function and moment equilibrium equations (at T11 to L5 levels) as equality constraints) then estimated muscle forces that counterbalanced computed required moment at each vertebral level (T1-T11 as a single rigid body, T12, L1, L2, L3, L4 and L5). To obtain physiologically valid muscle forces, we constrained muscle forces (F_i) to be greater than the passive force component (F_i^p) (Davis et al., 2003) and less than the sum of the passive force component plus the maximal active component:

$$F_i^p \leqslant F_i \leqslant F_i^p + PCSA_i\sigma_{max} \quad (1)$$

in which $PCSA_i$ and σ_{max} respectively denote physiological cross sectional area (of i^{th} muscle) and the upper bound of maximum muscle stress. To evaluate σ_{max} needed for convergence at each MVE task, we increased maximum muscle stresses (σ_{max}) starting from 0.2 MPa with the increment of 0.1 MPa. In this manner, required σ_{max} was calculated in each subject and each MVE task.

For subsequent comparison with recorded normalized EMGs in select muscles under a specific MVE of a subject, relative muscle activities were evaluated by normalizing their computed active forces ($F_i - F_i^p$) to their maximum active forces (${}^{max}F_i^a$) computed during all 6 MVE tasks:

$${}^{max}F_i^a = max\left({}^{F}F_i^a, {}^{E}F_i^a, {}^{RL}F_i^a, {}^{LL}F_i^a, {}^{RAx}F_i^a, {}^{LAx}F_i^a\right) = PCSA_i\sigma_i \quad (2)$$

where jF_i^a denotes computed active muscle force ($jF_i = jF_i^p + jF_i^a$) of i^{th} muscle at j^{th} task (i.e., extension (E), flexion (F), right lateral (RL), left lateral (LL), right axial (RAx) and left axial (LAx) MVEs). σ_i represents the peak muscle stress reached in i^{th} muscle under all MVEs of that subject. It should be noted that the use of foregoing σ_i ensures the appropriateness of comparisons between estimated and recorded relative muscle activities as a similar procedure was carried out when normalizing recorded EMGs.

Upper body gravity loads and their position were proportionally adjusted to the body weight and height, respectively, and partitioned along the spine (T1 to L5) (Pearsall et al., 1996) as well as arms, head-neck and hands (De Leva, 1996). The scaling algorithm adjusted both muscle architecture and passive spine responses based on imaging databases (Anderson et al., 2012; Shi et al., 2014) and biomechanical principles (Ghezelbash et al., 2016b). For more details on the model and the scaling algorithm see Fig. 2 and (Ghezelbash et al., 2016b). The nonlinear elastostatic analyses were carried out using ABAQUS (version 6.14, Simulia, Inc., Providence, RI, USA) finite element package program, and MATLAB (Optimization Toolbox) was used in the optimization algorithm.

At the interface between each subject and the dynamometer harness, identified visually in each task, equivalent forces (generating exactly the same moments recorded about the S1) were evaluated and applied on each model at respective contact points. During extension, flexion and lateral MVEs, contact points were located at the cranial-caudal heights situated respectively at the T8, T6 and shoulder joint and were offset out of the primary plane to generate measured moments (primary and coupled) about the S1. In the axial torque, the recorded axial MVE moment was applied at the T4. In addition and in order to reproduce accompanying coupled moments about the S1, required forces in the transverse plane were also calculated and applied to the T4.

To simulate the semi-seated posture of subjects during exertion tests and in accordance with the visual observation of subjects' configurations and radiological studies of individuals in seated posture, we reduced the lumbar lordosis (Bae et al., 2012; De Carvalho et al., 2010). Thus, we prescribed 9° (backward extension) and −13° (forward flexion) at the T11 in addition to 16° and 13°