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Identifying intrinsic and reflexive contributions to low-back stabilization

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A B S T R A C T

Motor control deficits have been suggested as potential cause and/or effect of a-specific chronic low-back pain and its recurrent behavior. Therefore, the goal of this study is to identify motor control in low-back stabilization by simultaneously quantifying the intrinsic and reflexive contributions. Upper body sway was evoked using continuous force perturbations at the trunk, while subjects performed a resist or relax task. Frequency response functions (FRFs) and coherences of the admittance (kinematics) and reflexes (sEMG) were obtained. In comparison with the relax task, the resist task resulted in a 61% decrease in admittance and a 73% increase in reflex gain below 1.1 Hz. Intrinsic and reflexive contributions were captured by a physiologically-based, neuromuscular model, including proprioceptive feedback from muscle spindles (position and velocity) and Golgi tendon organs (force). This model described on average 90% of the variance in kinematics and 39% of the variance in sEMG, while resulting parameter values were consistent over subjects.

1. Introduction

Low-back pain (LBP) is a common disorder, which affects 40–60% of the adult population annually in Western Europe and North America (Loney and Stratford, 1999; Picavet and Schouten, 2003). The effect of most treatments (e.g., anti-inflammatory drugs, neuromuscular training and cognitive therapy) is fairly small, and 60–75% of the patients have recurrent symptoms within a year with 10% developing chronic LBP (van den Hoogen et al., 1998). Motor control deficits (e.g., delayed 'reflex' responses, increased antagonistic co-contraction) have been suggested as potential cause and/or effect of LBP and its recurrent behavior (Cholewicki et al., 2000; Radebold et al., 2001; van Dieën et al., 2003).

Motor control provides an essential contribution to low-back stabilization, since the spine is inherently unstable without active musculature in spite of stiffness and damping provided by passive tissue (Bergmark, 1989; Crisco and Panjabi, 1991). The muscular contribution to stabilization of the spine involves muscle viscoelasticity and reflexive feedback. Muscle viscoelasticity comprises the stiffness and damping of the muscles and can be altered by co-contraction and selective muscle activity. Given the limited contribution of passive tissues especially in upright trunk postures and the difficulty to separate these components, properties of passive tissues and muscle viscoelasticity are usually lumped into intrinsic stiffness and damping. Feedback comprises visual, vestibular and proprioceptive contributions, where the latter is based on information of muscle length and muscle lengthening velocity from muscle spindles (MS) and on tendon force from Golgi tendon organs (GTO). Most studies on low-back stabilization have focused either on intrinsic stiffness and damping (e.g., Gardner-Morse and Stokes, 2001; Brown and McGill, 2009) or on reflexes (e.g., Radebold et al., 2001) by experimentally excluding the other component or analytically merging both. This could lead to incorrect estimates, especially because changes in co-contraction could result in changes in proprioceptive reflexes and vice versa (Matthews, 1986; Kirsch et al., 1993). Therefore, combined identification is essential, but only a few studies have pursued this for low-back stabilization.

Moorhouse and Granata (2007) and Hendershot et al. (2011) identified MS feedback and intrinsic stiffness of the trunk. However, low-back stabilization was not described, since their position-driven, upper-body perturbations stabilized the trunk. Goodworth & Peterka identified low-back stabilization focussing mainly on visual (Goodworth and Peterka, 2009) and vestibular (Goodworth and Peterka, 2010) feedback, while a simplified representation of proprioceptive reflexes (only stretch velocity MS feedback) and intrinsic contributions (only stiffness) was used. Thus, a detailed analysis of the contribution of proprioceptive reflexes to low-back stabilization is still lacking.

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The goal of this study was to simultaneously identify intrinsic and reflexive contributions to low-back stabilization in healthy subjects. This approach could help identify motor control deficits in LBP.

2. Methods

2.1. Subjects

Fifteen healthy adults (age, 23–58 year; mean age, 35 year) participated in this study and gave informed consent according to the guidelines of the ethical committee of VU University Amsterdam. Subjects did not experience LBP in the year prior to the experiments.

2.2. Experiments

During the experiments, subjects assumed a kneeling-seated posture, while being restrained at the pelvis (Fig. 1). A force perturbation $F_{pert}(t)$ was applied in ventral direction at the T10-level of the spine by a magnetically driven linear actuator (Servotube STB2510S Forcer and Thrustrod TRB25-1380, Copley Controls, USA). For comfort and better force transfer, a thermoplastic patch (4×4 cm) was placed between the actuator and the back of the subject. To reduce the effects of head and arm movement during the measurements, the subjects were instructed to place their hands on their head.

Visual feedback depicting the trunk rotation in sagittal (flexion/extension) and coronal (lateral bending) plane was provided to the subjects. Task instructions were to minimize the flexion/extension excursions (*Resist task*), or to relax as much as possible while limiting flexion/extension to about 15 degrees (*Relax task*). In addition, subjects were instructed in both tasks to minimize lateral flexion. Both tasks were repeated four times with the same perturbation signal.

The perturbation $F_{pert}(t)$ (Fig. 2) consisted of a dynamic disturbance of ± 35 N combined with a 60 N baseline preload to maintain contact with the subject, because the actuator was not connected to the subject and therefore only capable of pushing. The dynamic disturbance (Fig. 2) was a crested multisine signal (Pintelon and Schoukens, 2001) of 20 sec duration with 18 paired frequencies, which were logarithmically distributed within a bandwidth of 0.2–15 Hz. To reduce adaptive behavior to high frequent perturbation content, the power above 4 Hz was reduced to 40% (Mugge et al., 2007). Because the perturbation was random-appearing, subjects were not expected to react with voluntary activation on the perturbation.

Each run consisted of a ramp force increase to preload level (3 s), a stationary preload (2 s), a start-up period to reduce transient behavior (the last 5 s of the dynamic disturbance), and twice the dynamic disturbance (2×20 s), which resulted in 50 s per run.

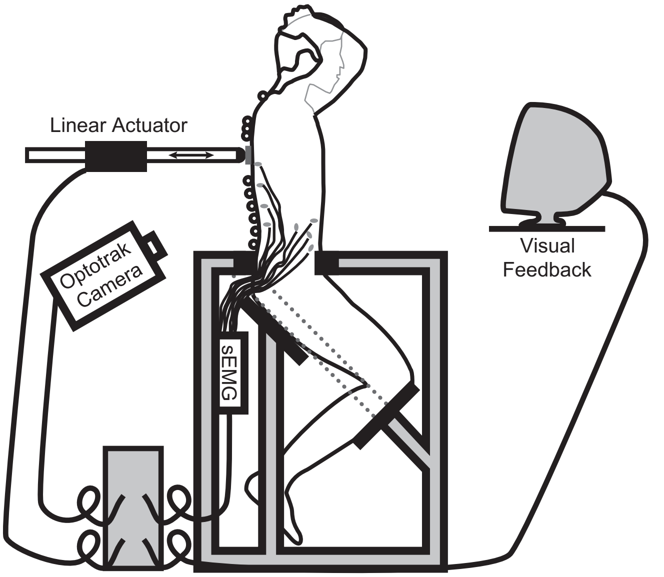

Fig. 1. Experimental setup. Subjects were restrained at the pelvis and positioned in a kneeling-seated posture, while Optotrak markers (○) and EMG electrodes are attached.

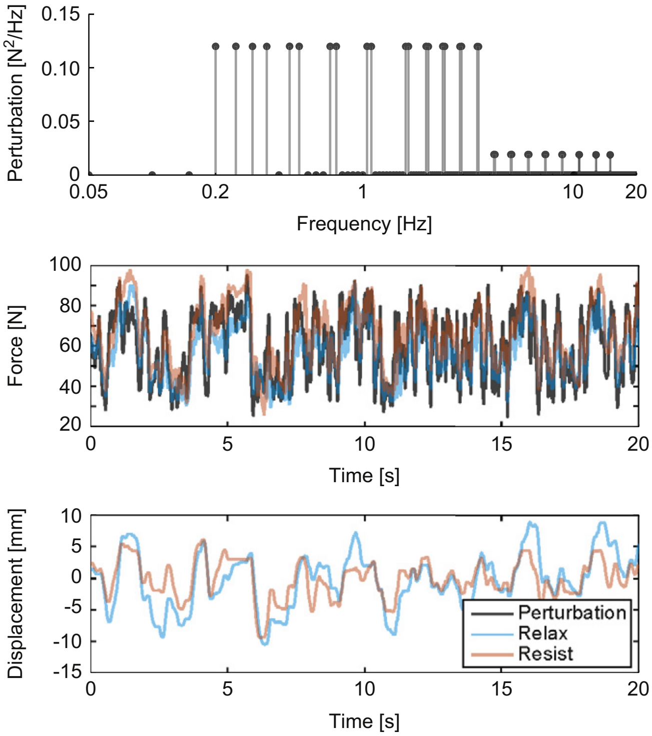

Fig. 2. The force perturbation F_{pert} (black) is projected in frequency domain (TOP) and time domain (MIDDLE). The resulting contact forces $F_C(t)$ (MIDDLE) and actuator displacements $x_A(t)$ (BOTTOM) are shown in time domain during a relax task (blue) and a resist task (red). (For interpretation of the references to color in this figure legend, the reader is referred to the web version of this article.)

2.3. Data recording and processing

Kinematics of the lumbar vertebrae (L1–L5), the thorax (T1, a cluster of markers at T6, T12), and the pelvic restraint were measured using 3D motion tracking at 100 Hz (Optotrak3020, Northern Digital Inc, Canada). The trunk rotation angle (based on markers at T12 and the pelvic restraint) in sagittal and coronal plane was provided as visual feedback to the subjects in real-time. The actuator displacement $x_A(t)$ and contact force $F_c(t)$ between the rod and the subject were measured at 2000 Hz (Servotube position sensor & Force sensor FS6-500, AMTI, USA). Trunk kinematics were described in terms of translation, since kinematic analysis indicated that an effective low-back bending rotation point, necessary to define rotations, was not well defined and inconsistent over subjects and tasks. Activity of sixteen muscles (8 bilateral pairs as listed in Table 1) was measured at 1000 Hz (surface electromyography (sEMG) Porti 17, TMSi, the Netherlands) as described in Willigenburg et al. (2010). The EMG data $e_j(t)$ (with $j = \sharp$muscle) was digitally filtered (zero-phase, first-order, high-pass) at 250 Hz (Staudenmann et al., 2007) and then rectified.

All fifteen subjects showed a comparable admittance with an actuator displacement *rms* of 2.72±0.49 mm (relax) and 1.78±0.36 mm (resist). Further analysis of local low-back bending patterns (van Drunen et al., 2012) showed substantial low-back bending in eight subjects where at least 32% of the trunk rotations were attributed to bending above L5 (while measurements were not below L5) during both task instructions. In the other seven subjects, at least one task instruction resulted in less than 6% trunk rotation attributed to bending above L5, suggesting that bending below L5 and/or pelvic rotations accounted for much of the observed trunk rotations. Hence, the data collected on these subjects was not suitable for studying lumbar stabilization. Therefore, this paper will consider only the eight subjects demonstrating substantial low-back bending.

2.4. System identification

Closed loop system identification techniques (van der Helm et al., 2002; Schouten et al., 2008a) were used to estimate the translational low-back admittance ($\hat{H}_{adm}(f)$) and reflexes ($\hat{H}_{emg}(f)$) as frequency response functions (FRFs). The admittance describes the actuator displacement ($x_A(t)$) as a function of the contact force ($F_c(t)$), representing the inverse of low-back mechanical impedance. The reflexes describe the EMG data ($e_j(t)$) as a function of the actuator displacement ($x_A(t)$). Because the subjects interacted with the actuator, FRFs were estimated