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Spinal loads and trunk muscles forces during level walking – A combined *in vivo* and *in silico* study on six subjects

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ABSTRACT

During level walking, lumbar spine is subjected to cyclic movements and intricate loading of the spinal discs and trunk musculature. This study aimed to estimate the spinal loads (T12–S1) and trunk muscles forces during a complete gait cycle.

Six men, 24–33 years walk barefoot at self-selected speed (4–5 km/h). 3D kinematics and ground reaction forces were recorded using a motion capturing system and two force plates, implemented in an inverse dynamic musculoskeletal model to predict the spinal loads and trunk muscles forces. Additionally, the sensitivity of the intra-abdominal pressure and lumbar segment rotational stiffness was investigated.

Peak spinal loads and trunk muscle forces were between the gait instances of heel strike and toe off. In L4–L5 segment, sensitivity analysis showed that average peak compressive, antero-posterior and medio-lateral shear forces were 130–179%, 2–15% and 1–6%, with max standard deviation (±STD) of 40%, 6% and 3% of the body weight. Average peak global muscles forces were 24–55% (longissimus thoracis), 11–23% (iliocostalis thoracis), 12–16% (external oblique), 17–25% (internal oblique) and 0–8% (rectus abdominus) of body weight whereas, the average peak local muscles forces were 11–19% (longissimus lumborum), 14–31% (iliocostalis lumborum) and 12–17% (multifidus). Maximum ± STD of the global and local muscles forces were 13% and 8% of the body weight.

Large inter-individual differences were found in peak compressive and trunk muscles forces whereas the sensitivity analysis also showed a substantial variation.

1. Introduction

Level walking is a relevant locomotion in daily life. An average walking speed of 5 km/h was found among the general population (Browning et al., 2006). Normal walking is sometimes recommended for post-operative rehabilitation, less likely to cause activity related spinal injury. The underlying mechanics to produce forward locomotion is intrinsically complicated, comprising neuromuscular architecture driving the skeletal system to provide the desired locomotion. The active (muscles) and passive structures (e.g., spinal disc, ligaments and tendons) coordinate in forward locomotion to attain the dynamic equilibrium. From mechanical perspective, it is important to understand the motion-loading relationship, especially the pattern of spinal loading and corresponding trunk muscles forces during a gait cycle, to make proper diagnosis for pathological situations and recommend proper treatment.

During level walking (1–5 km/h), *in vivo* measurements of the spinal loading in telemetrized vertebral body replacements showed that resultant forces were between 140 and 370% relative to standing (%STG) (Rohlmann et al., 2014, 2013). Alternatively, intradiscal pressure measurements revealed an increase of loading between 106 and 130 (%STG) (Wilke et al., 1999). Also, electromyographic (EMG) measurements showed that peak muscles activities occurred mostly during the gait instance of double support (i.e., in between heel strike and contralateral toe off) (Anders et al., 2007; Arendt-Nielsen et al., 1996; Callaghan, 1999; Cappozzo, 1984; Ceccato et al., 2009; Cromwell et al., 1989; Lamoth et al., 2006a, 2006b; Thorstensson et al., 1982; Van der Hulst et al., 2010).

Some combined *in vivo* and *in silico* studies used EMG measurements and musculoskeletal models to predict the spinal loads

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(L3–L4 and L4–L5) and trunk muscles forces (Callaghan, 1999; Cappozzo, 1984), where maximum compressive loads in a lumbar motion segment was predicted between 1.1 and 2.6 times the body weight (BW) at 4–5 km/h. Other studies predicted and compared the spinal loads and trunk muscle forces between normal and lower extremity amputees (Hendershot and Wolf, 2014; Shojaei et al., 2016; Yoder et al., 2015), where they showed that the compressive forces in a lumbar segment could increase up to 2.6 times the BW.

While using musculoskeletal models, parameters such as intra-abdominal pressure (IAP) or lumbar segment rotational stiffness (SRS), could influence predicted spinal loads and trunk muscles forces, however, they were not investigated before. During level walking, *in vivo* measurements showed that IAP is cyclic (Grillner et al., 1978; Shaw et al., 2014), having peak values coincident with ground reaction forces, which could reduce the spinal loads and trunk muscles forces, whereas the reduction in SRS could lead to higher spinal loads and forces in certain trunk muscles groups.

Therefore, the aim of this study was to estimate the spinal loads and trunk muscles forces in young subjects for barefoot level walking at normal speed, show inter-individual differences and sensitivity to the modeling choices such as the inclusion of the IAP and lumbar SRS.

2. Methods

2.1. Study protocol

Six male subjects, 24–33 years were selected at the Julius Wolff Institut, Charité–Universitätsmedizin Berlin (Table 1). Few exclusion criteria were considered: (1) Body Mass Index (BMI) above 26 kg/m^2, (2) hip or low back pain persisting for more than 6 weeks and (3) musculoskeletal disorders influencing normal kinematics of the spine and lower extremities. All participants were explained about the procedure and then a written consent was made for the participation in these measurements.

2.2. Measurement apparatus

A 3D motion capturing and analysis system (VICON Motion Systems, Inc, Oxford, United Kingdom) was used, having ten cameras operating at 150 Hz. Two force plates (AMTI, MA, USA), embedded in the laboratory floor recorded ground reaction forces which were sampled at a frame rate of 900 Hz in synchronization with the cameras.

2.3. Subject preparation and in vivo measurement

Forty-seven reflective markers (12 mm diameter) were placed on the anatomical landmarks per customized full-body skin marker set (Fig. 1). Six markers were placed on the superior spinal processes of the five lumbar vertebrae (L1–L5) and the sacrum (S1).

For each subject, six trails were recorded. With the aim of avoiding any constraints on the movement, no specific instructions were given. Each subject walked barefoot at self-selected speed along a straight 10-m walkway with natural arm swing. The average walking speed measured in these trials was 4–5 km/h (Table 1). A gait trial was considered acceptable only when the entire foot cleanly struck the force platform.

Further, motion capture data was pre-processed with the Nexus software. A zero-lag second order low pass Butterworth filter was used. For marker trajectories and force data, the cut-off frequencies were 7 Hz and 23 Hz respectively, whereas missing or occluded markers were reconstructed using (1) the spline fill, (2) the pattern fill or (3) the rigid body algorithm (VICON, 2016).

2.4. Musculoskeletal model

A full body musculoskeletal model (AnyBody, v.1.6.2.) was used which included the Twente Lower Extremity Model (TLEM) (Klein Horsman et al., 2007) and detailed lumbar spine model (De Zee et al., 2007). The trunk muscles were grouped as global and local, according to Table 2 (El-Rich et al., 2004).

The pre-processed motion capture data were used as a kinematic input for the model. Optical markers placed on the lumbar spine captured only an overall 3D motion. Therefore, a predefined distribution (lumbar spine rhythm) of 3D rotations (L1–L2:33.0%, L2–L3:26.5%, L3–L4:20%, L4–L5:13.5%, L5–S1:7%) at each lumbar segment was used in the model. The lumbar spine rhythm defined increasing 3D rotations from lower to upper lumbar segments, consistent with *in vivo* measurements of the lumbar spine kinematics during level walking (Gombatto et al., 2015).

In the model, SRS in flexion, extension, lateral bending and axial rotation was considered (Schmidt et al., 1998) and ligaments were not modelled separately to simplify the modeling approach.

In AnyBody model, the IAP was generated due to the change in abdominal volume. The abdominal volume was wrapped around by the transverse abdominus as well as surrounded by internal oblique, external oblique and rectus abdominus muscles. The IAP was generated by an artificial muscle acting on the abdominal volume. During level walking, the posture or the body movement change the volume measure that lead to the generation of IAP, which effects the lumbar spine segments anteriorly via kinetic attachment points. This influences the overall muscle recruitment and therefore the predicted joint reactions and muscles forces. The maximum IAP that could be generated by the model was limited to an upper bound value of 26.6 kPa (Essendrop, 2003).

To investigate the sensitivity of the spinal loads and trunk muscles forces to IAP and lumbar SRS, the Basic model (without IAP and SRS) was used with four different parameter settings (Basic, Basic + IAP, Basic + IAP + SRS and Basic − IAP + SRS). A total of 48 simulations were performed with four models, two trials and six subjects, whereas results of 46 simulations were presented.

Table 1
Six subjects without low back pain. BM: body mass, BH: body height, STD: standard deviation.

Subject (male = 6)	BM (kg)	BH (cm)	Ave. steps per min	Ave. speed (km/h)
1	78.90	177	102	4.46
2	68.20	176	110	5.08
3	73.30	181	108	5.29
4	83.90	186	110	5.36
5	82.50	186	106	4.93
6	63.90	176	100	4.46
Average ± STD	75.0 ± 8.03	180 ± 4.76	106 ± 4.19	4.93 ± 0.39