Effects of Sex, Age, Body Height and Body Weight on Spinal Loads: Sensitivity Analyses in a Subject-Specific Trunk Musculoskeletal Model

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ABSTRACT

Subject-specific parameters influence spinal loads and the risk of back disorders but their relative effects are not well understood. The objective of this study is to investigate the effects of changes in age (35-60 years), sex (male, female), body height (BH: 150-190 cm) and body weight (BW: 50-120 kg) on spinal loads in a full factorial simulation using a personalized (spine kinematics, geometry, musculature and passive properties) kinematics driven musculoskeletal trunk finite element model. Segmental weight distribution (magnitude and location along the trunk) was estimated by a novel technique to accurately represent obesity. Five symmetric sagittal loading conditions were considered, and main effect plots and analyses of variance were employed to identify influential parameters. In all 5 tasks simulated, BW (98.9% in compression and 96.1% in shear) had the greatest effect on spinal loads at the L4-L5 and L5-S1 levels followed by sex (0.7% in compression and 2.1% in shear), BH (0.4% in compression and 1.5% in shear) and finally age (<5.4%). At identical BH and BW, spinal loads in females were slightly greater than those in males by ~4.7% in compression and ~8.7% in shear. In tasks with no loads in hands, BW-normalized spinal loads further increased with BW highlighting the exponential increase in spinal loads with BW that indicates the greater risk of back disorders especially in obese individuals. Uneven distribution of weight in obese subjects, with more 17 BW placed at the lower trunk, further (though slightly <7.5%) increased spinal loads.

Keywords: Obesity, Age, Sex, Body Height, Trunk Biomechanical Modeling, Finite Element

1 INTRODUCTION

Back pain is a prevalent health issue worldwide (Hoy et al., 2014; Hoy et al., 2010b) with significant social and economic burdens on individuals and society (Deyo et al., 1991; Katz, 2006; Rapoport et al., 2004). Ageing (Hoy et al., 2012), obesity (Deyo and Bass, 1989) and body height (BH) (Leclerc et al., 2003) are recognized as risk factors. Ageing, for instance, increases the prevalence of back pain and alters its etiology (DePalma et al., 2011; Hoy et al., 2012; Hoy et al., 2010a). While back pain in younger individuals has often discogenic origins, it is in older individuals mainly from facets and sacroiliac joint (DePalma et al., 2011; DePalma et al., 2012; Dionne et al., 2006). As a rising global health problem (Flegal et al., 2012; Wang et al., 2011), obesity has also been associated with back pain (Deyo and Bass, 1989; Heuch et al., 2010; Koyanagi et al., 2015; Leboeuf–Yde et al., 1999; Shiri et al., 2009; Shiri et al., 2014; Smuck et al., 2014; Webb et al., 2003). These studies define obesity based on body mass index (BMI) whereas waist to hip ratio (Han et al., 1997; Yip et al., 2001), waist circumference (Lean et al., 1998; Shiri et al., 2013; Taanila et al., 2012) and body weight (BW) (Croft and Rigby, 1994; Heuch et al., 2015b) have also been used. As a risk factor, greater BH can also cause back pain in females (Heuch et al., 2015a; Yip et al., 2001), males (Walsh et al., 1991) or both (Hershkovich et al., 2013). Though some studies question the likely role of BH (Han et al., 1997), others suggest that taller stature could predispose individuals to back pain (Coeuret-Pellicer et al., 2010). Correlation between gender and back pain has been reported (DePalma et al., 2012; Schneider et al., 2006). Though personalized factors have been indicated in back pain, underlying mechanisms remain yet unknown.

The above factors likely alter spinal loads. To estimate loads on spine, *in vivo* studies, though valuable, are costly, limited and invasive (Dreischarf et al., 2015; Rohlmann et al., 2013; Sato et al., 1999). Musculoskeletal models have emerged as robust and relatively accurate alternatives. Hajihosseinali et al. (2015) applied an image-based anisotropic scaling method to modify musculature morphology in a musculoskeletal trunk model while investigating the effects of changes in BW on spinal loads. They reported that BW substantially influences spinal loads particularly at flexed postures. Using a linear and isotropic scaling scheme in AnyBody Modelling System (Damsgaard et al., 2006), Han et al. (2013) found that the spinal shear and compression forces change linearly with BH and BW though the effect of BW is more pronounced. To investigated age-related hyperkyphosis by a static model of the spine, Bruno et al. (2012) considered three spinal configurations (hyperkyphosis alone, with pelvic tilt or with increased lordosis) and reported that changes in both kyphosis and spinal posture affect spinal loads. Nevertheless, to-date no study has comprehensively investigated the likely effects of all subject-specific parameters 52 of age, sex, BH and BW on spinal loads.

Computing spinal forces by multi-joint trunk musculoskeletal models, especially when BW changes, requires an accurate segmental weight distribution along the spine (T1 to L5). Pearsall et al. (Pearsall, 1994; Pearsall et al., 1996) evaluated this distribution in lean individuals using CT imaging. For overweight and obese individuals, however, available studies have estimated only the total trunk mass center by MR images (Matrangola et al., 2008), X-ray absorptiometry scans (Chambers et al., 2010) and 3D body scans (Pryce and Kriellaars, 2014). Consequently, the required segmental weight distribution in overweight and obese individuals has not yet been estimated.

We aim to comprehensively investigate the effects of alterations in age, sex, BH and BW on spinal loads. To adequately account for the overweight and obese individuals, we initially develop a novel technique to estimate segmental weight distribution along the trunk (T1 to L5) as BW alters. Moreover, using an updated validated nonlinear finite element (FE) subject-specific trunk musculoskeletal model (Ghezelbash et al., 2016) in conjunction with personalized spinal kinematics (with respect to age and sex) (Pries et al., 2015), we evaluate spinal loads and sensitivities therein as individual parameters alter in a full factorial simulation (90 cases) taking 4 independent factors (age, sex, BH and BW) in five sagittally symmetric tasks. In accordance with earlier studies, we hypothesize that spinal loads are much more sensitive to variations in BW than in sex, BH and age.

2 METHODS

2.1 Musculoskeletal Model of Trunk

The development and validation of a nonlinear FE, subject-specific, musculoskeletal model of the trunk for symmetric-asymmetric tasks are reported elsewhere (Ghezelbash et al., 2016). The model includes a comprehensive sagittally-symmetric muscle architecture (126 muscle fascicles) and spinal motion segments (T11-T12 to L5-S1) that are simulated as shear-deformable beam elements with nonlinear properties (Shahvarpour et al., 2016; Shirazi-Adl, 2006). To estimate muscle forces, the musculoskeletal trunk model is driven by measured kinematics while minimizing sum of squared muscle stresses (Arjmand et al., 2010; Arjmand and Shirazi-Adl, 2006). Moreover, to adjust the model in accordance with subject's personal parameters (age, sex, BH and BW), we use a physiological-based scaling method that modifies both muscle architecture (geometry and area of muscles) and passive joint properties in accordance with imaging studies (Anderson et al., 2012; Shi et al., 2014) and biomechanical principles (Ghezelbash et al., 2016).

2.2 Body Weight Distribution

For BMI<25 kg/m^2 , upper trunk BW (head, arms and trunk) is distributed based on the literature (De Leva, 1996; Pearsall, 1994) and similar to our earlier works (Arjmand et al., 2010; El Ouaaid et al., 2015; Shahvarpour et al., 2015). However, for BMI>25 kg/m^2 , a new approach described below, is developed since the existing data, collected on lean individuals, cannot accurately be extended to obese and overweight ones.

3D Reconstruction from 2D images: 3D body shapes of subjects are initially reconstructed (e.g., by spline curves) using available 2D image datasets of thousands of human laser scans (BMI Visualizer, Perceiving Systems Department, Max Planck Institute for Intelligent Systems, Germany) in the sagittal and frontal planes (Allison et al., 2013). BW of each reconstructed 3D body is then estimated and calibrated based on a reported regression equation (Velardo and Dugelay, 2010).

Placement of the vertebral column: We employ a standing MR image of a male subject (BMI=26 kg/m^2) produced from data acquired in a previous study (Meakin et al., 2008b) including images of the lumbar and thoracic spines to position the spinal column within foregoing body images by fitting the boundaries of the MR and body scan images. With feet and head fixed as landmarks (Fig. 1), it is assumed that the spine preserves its relative position to landmarks as BMI varies. To validate the 3D reconstruction and this positioning algorithm of the spine within images, estimated segmental masses and mass centers as well as the total trunk mass in lean subjects are subsequently compared to those reported in supine based on CT images (Pearsall, 1994; Pearsall et al., 1996).

Calculation of mass centers for obese and overweight subjects: For these subjects (BMI>25 kg/m^2), BW is initially partitioned into two parts:

$$BW = BW_R + BW_A, \qquad (1)$$

where is the reference body weight assuming a BMI=25 kg/m^2 and is the additional body weight. While the reference body weight (BW_R) is distributed in accordance with the available data for lean subjects (De Leva, 1996; Pearsall, 1994), the additional body weight (BW_A) is assumed to be made of adipose tissue (yellow areas in Fig. 1) with distribution based on reconstructed 3D shapes. This procedure is carried out at BMI=25, 30, 35 and 40 kg/m^2 at BH=173 cm and then scaled for other BHs. Segmental masses of adipose tissue are scaled proportional to BW_A, and the mass centers are adjusted proportional to BH. Additional masses of the head and arms are estimated from the reconstructed 3D surface images with mass center locations reported in the literature (De Leva, 1996). To evaluate the validity of foregoing partitioning approach, we compared the estimated density of the additional material (yellow areas in Fig. 1) with the reported density of adipose 113 tissue (919.6 kg/m^3 (Farvid et al., 2005)).

Sensitivity analysis: The forgoing method of estimating weight along the spine applies to the general obese population. Therefore, to investigate the likely effects of extreme weight distributions on spinal loads at BMI=35 kg/m^2, BH=173 cm and age=47.5 years, we replace segmental weights at lumbar levels with those of either 30 kg/m^2 (less weight around the waist) or 40 kg/m^2 (more weight around the waist) while maintaining the BW constant by proportionally adjusting segmental weights in the upper thorax region. Flexion at 20^o and 50^o are simulated with no load in hands.

2.3 Full Factorial Design

A full factorial simulation with 4 independent factors (2 sexes x 3 ages x 3 BHs x 5 BWs=90 cases, Table 1) is considered. All 90 cases are simulated under five sagittally symmetric tasks (in total 450 simulations): 1- upright standing holding 5 kg in hands anteriorly, 2- and 3- trunk flexion at 20^o and 50^o with no load in hands, 4- flexion at 20^o with 10 kg in hands, and 5- flexion at 50^o with 5 kg in hands. In each analysis, initially the reference subject-specific upright standing posture under gravity alone is sought from corresponding personalized undeformed (unloaded) configuration by a moment optimization approach in which the sum of sagittal moments at the T11-L5 levels is minimized under upper body gravity loads (Shirazi-Adl et al., 2002). Subsequently, in each flexed posture, total thoracolumbar (T11-L5) and sacral (S1) rotations are determined in accordance with reported measured lumbopelvic ratios that are personalized for sex and age (Pries et al., 2015). Thoracolumbar (T11-S1) rotations in each task are partitioned between its motion segments; 6.0% at the T11-T12, 10.9% at the T12-L1, 14.1% at the L1-L2, 13.2% at the L2-L3, 16.9% at the L3-L4, 20.1% at the L4-L5, and finally 18.7% at the L5-S1 (Gercek et al., 2008; Ghezelbash et al., 2016; Hajibozorgi and Arjmand, 2015).

Statistical analysis: Main effect plots are utilized to investigate the effects of various factors and analyses of variance (ANOVA) are carried out to determine relative importance of various factors (Dar et al., 2002; Meakin et al., 2003). Each response (shear and compression forces at the L4-L5 and L5-S1 levels) are considered separately with interactions neglected (reduced order model). It should be noted that to determine the contribution of each factor in the response, we use sums of squares (not p-values), so there is no restriction in using ANOVA even if our musculoskeletal model is deterministic.

5 3 RESULTS

Body weight distribution: For the reference case (BMI=25 kg/m^2), the correlation coefficient (Pearson's r) and root-mean-square error (RMSE) between the predicted (as described earlier in Methods) and reported (Pearsall, 1994) locations of the segmental mass center at various spinal levels are 0.92 and 6.8 mm, respectively. The correlation coefficient and RMSE for segmental weights (from T1 to L5) are 0.83 and 3.2 N, respectively. Besides, the absolute relative error between the predicted and reported (Pearsall, 1994) whole trunk mass is 9.0%. Low errors in combination with high correlation coefficients indicate the relative accuracy in our body weight distributions and positioning of the spine in each case within the reconstructed personalized body shape images.

In overweight and obese cases (BMI=30, 35 and 40 kg/m^2), mass center locations of the additional weights are found relatively close to each other (**Fig. 2**a). Additional segmental weights vary along the spine (**Fig. 2**b). Absolute relative error of the predicted density of the adipose tissue for different BHs (150–190 cm) and BMIs (30-40 kg/m^2) is 6.6 ± 3.8% (mean± standard deviation).

Spinal loads: Changes in BW, sex and BH influence spinal loads. Increasing BW from 55 to 120 kg nearly doubles compression forces at the L4-L5 and L5-S1 levels (**Figs 3** and 4). Based on the results of all simulations and under identical parameters, females experience slightly larger (~4.7% in compression and ~8.7% in shear) loads than males. With no load in hands, BW-normalized spinal loads increase with BW (**Fig. 5**). However, these trends reverse when a load is added in hands (**Fig. 5**). BH-normalized spinal loads drop linearly with BH for all loading conditions (Fig. 6), except for the largest BW at lower BH values. Loads in females are slightly larger than their male counterparts. Also, sensitivity analyses on weight distribution of obese individuals at BMI=35 kg/m^2, BH=173 cm and age=47.5 years show relatively small differences in spinal loads (peaks of 7.5% in shear and 6.2% in compression).

Statistical analysis: Main effect plots (for all analyses, **Fig. 7**) reveal that the spinal loads at the L5- S1 increase nearly proportionally with BW. For identical age, BW and BH, males have lower spinal loads than females with larger differences in shear than in compression. BH and age, on the other hand, do not noticeably affect spinal loads. Similar trends are obtained for spinal loads at the L4-L5 disc. According to the analyses of variance (Table 2), BW (~98.9% for compression and 96.1% for shear) is the main contributing factor while sex (~0.7% for compression and 2.1% for shear), BH (~0.4% for compression and 1.5% for shear) and age (<5.4%) have much less effects.

4 DISCUSSION

In this study, we investigated the sensitivity of spinal loads at the L4-L5 and L5-S1 levels to changes in personalized factors (age, sex, BW and BH) under 5 different sagittal-symmetric loads. Proper accounts of gravity distribution along the spine in obese subjects and of initial posture of spine under gravity were made. In confirmation of our hypothesis, changes in BW (means of 98.9% in compression and 96.1% in shear) influenced to a great extent the spinal loads whereas the role of sex (0.7% in compression and 2.1% in shear), BH (0.4% in compression and 1.5% in shear) and age (<5.4%) (Table 2) were much smaller. In comparison to males, females (at identical age, BH and BW) experienced greater spinal loads (~4.7% in compression and ~8.7% in shear).

4.1 Model Evaluation

Novelties: Segmental weight distribution for overweight and obese individuals was presented in this study for the first time. It was also applied (utilizing a subject-specific model of the trunk with scaled muscle geometry and passive joint properties along with personalized kinematics) to the investigation of the effects of age, sex, BH and BW on spinal loads. The initial posture in upright standing under gravity alone was also personalized for each subject. **Shortcomings:** Apart from limited *in vivo* studies available for validation (Dreischarf et al., 2016) and the limitations (e.g., neglecting intra-abdominal pressure and coactivity) noted elsewhere (Arjmand et al., 2006; El-Rich et al., 2004; Ghezelbash et al., 2015; Ghezelbash et al., 2016), the model did not converge in taller, older, obese and female individuals with load in hands (≥5 kg) at large flexion angles (≥70º) since in contrast to spinal loads, the contribution of BH to the required moments is not negligible, and muscles in females could not counterbalance the induced required moments. In accordance with the input data used (Anderson et al., 2012; Pries et al., 2015; Shi et al., 2014) and to avoid spinal disorders observed at older ages, we chose the range of 35-60 years that also covers active working ages. The proposed method of estimating segmental weights involves some simplifying assumptions: reconstruction of the 3D body from 2D images (Allison et al., 2013), placement of the spine in the reconstructed body, homogenous distribution of density (Pryce and Kriellaars, 2014), scaling weight distribution of adipose tissue and extrapolation of weight distribution for three cases (BH=150 cm at BMI=46.1 and 53.3 kg/m^2; BH=170 cm at BMI=41.5 kg/m^2) due to limitations in the available database (BMI Visualizer, Perceiving Systems Department, Max Planck Institute for Intelligent Systems, Germany). Nonetheless, the method yielded results in satisfactory agreement with the distribution of gravity loads along the spine in lean subjects (Pearsall, 1994; Pearsall et al., 1996) and the density of additional adipose tissue (Farvid et al., 2005). Moreover, for an extreme underweight case (BMI=15.2 kg/m^2), we used the mass 204 distribution similar to normal weight individuals. Only sagittally symmetric tasks were simulated here; the sensitivity of spinal loads to personalized factors may alter in asymmetric tasks although conclusions likely remain unchanged. Changes in the thoracic kyphosis angle or alterations in the initial lordosis (Meakin et al., 2008a) could influence results.

4.2 Comparisons

Age: No study has investigated changes in age while employing a detailed trunk musculoskeletal model; existing studies (Boocock et al., 2015; Shojaei et al., 2016; Song and Qu, 2014) employed link-segment models (for limitations and shortcomings see Rajaee et al., 2015). In contrast to our findings, Shojaei et al. (2016) reported significantly lower shear force at the L5-S1 in younger participants. The use of a dynamic link-segment model with no muscles and passive spine, different anthropometric input data and age groups (22-68 years versus 35-60 years) as well as asymmetry in tasks could be some likely sources for different findings.

Sex: Marras et al. (1995, 2003) estimated lower spinal loads in females (except the anterior-posterior shear at the L5-S1 disc (Marras et al., 1995)). In these studies, however, male participants were, in average, heavier and taller than females. Differences in BH and BW along with using a single level EMG-driven model without a comprehensive scaling algorithm (Dreischarf et al., 2016) could have played a role in lower estimation of spinal loads in females. In agreement with our findings, Shojaei et al. (2016) computed larger shear forces (~6%) in females.

BW & BH: In accordance with our findings, Hajihosseinali et al. (2015) and Han et al. (2013) found also that BW markedly affects spinal loads and Han et al. (2013) reported that BH has less effects on spinal loads. Moreover, obese individuals experience more spinal shrinkage (Yar, 2008) and implant subsidence (Behrbalk et al., 2013).

4.3 Interpretations

Among various parameters, spinal loads are particularly sensitive to passive joint properties, muscle moment arms, lumbopelvic ratio and net external moments. Greater passive joint stiffness as well as muscle moment arms markedly reduce muscle forces and consequently spinal loads. Likewise, increasing the lumbopelvic ratio (at a given posture) reduces spinal loads by accentuating the load-carrying role of the passive spine (Tafazzol et al., 2014). Finally, greater net moment at a spinal level, being due to larger external/gravity load or changes in the posture, tends to increase muscle forces and spinal loads. Subject specific factors affect spinal loads by altering the forgoing parameters. For instance, at identical age, BH and BW, female spines experience greater loads due to associated smaller muscle moment arms and passive joint contributions (Table 3). Results indicate that changes in age hardly influence spinal loads that could be due to opposing trends in the lumbopelvic ratio and passive contributions/muscle moment arms (Table 3). By increasing external moments, BW markedly influence spinal loads while the increase in external moments due to BH is almost counterbalanced by the larger muscle moment arms and passive joint contributions (Table 3).

With no load in hands, the BW-normalized spinal loads further increase with BW in both sexes and particularly at higher BHs (Fig. 5). This highlights the accentuating role of BW in increasing spinal loads especially in obese individuals. It further demonstrates the exponential increase in spinal loads with BW. This trend, however, disappears in conditions with a load in hands (Fig. 5) with larger decreases under greater loads in hands which indicates the substantial effect of external loads on spinal forces. BH-normalized spinal loads decreased in all loading conditions with BH for all BWs (except and in particular in shear forces at larger BWs and smaller BHs, Fig. 6). This drop in BH-normalized spinal forces when compared to the opposite increase in BW-normalized loads versus BW further suggests the important role of BW on spinal loads.

Due to the controversial relation between the lumbar lordosis and age, sex, BH and BW (or alternatively BMI) (Been and Kalichman, 2014), the initial (undeformed) lumbar lordosis was kept constant in all models. Some studies found no association between the lordosis and age (Kalichman et al., 2011; Murrie et al., 2003) while others reported age-related decrease (Amonoo-Kuofi, 1992) or increase (Tüzün et al., 1999) in the lordosis. In a preliminary study (not reported here) simulating forward flexion, however, the initial L1-S1 lordosis Cobb's angle was increased by 10^{o} (from 46^{o}) while preserving the kyphosis angle and the sacral plumb line in agreement with the literature (Endo et al., 2010; Jackson et al., 1998; Park et al., 2013). Results demonstrated that it primarily influenced the relative ratio of shear and compression forces especially at the L5-S1 under larger flexion angles. Factors such as age and sex could, therefore, potentially influence spinal loads indirectly through alterations in lordosis.

Excessive spinal loads have been recognized as a risk factor of back pain (Bovenzi et al., 2015; Coenen et al., 2013; Marras et al., 1995) and disc degeneration (Adams et al., 2000; Rannou et al., 2004; Stokes and Iatridis, 2004). Thus, greater BWs that yield larger (BW normalized) spinal loads and fat accumulation at the (upper) trunk in comparison with more weight around and below the waist can predispose individuals to higher risk of back disorders. It is to be noted that consideration of loads when normalized to subject's BW is more appropriate since unlike absolute loads, BW-normalized loads to some extent automatically take account of anthropometric differences. Overall, increase in spinal loads (Singh et al., 2015), reduction in postural stability (Corbeil et al., 2001; Hue et al., 2007), fall due to slipping (Allin et al., 2016), limitations in goal-oriented movements (Berrigan et al., 2006) and metabolic changes (Samartzis et al., 2013) due to obesity increase back pain risk of injury while some alterations such as the associated decrease in the ranges of motion (Park et al., 2010; Vismara et al., 2010), adaptation (Porter et al., 1989), physical activity (Smuck et al., 2014) and the unloading role of IAP (due to increased diaphragm area (Shi et al., 2014)) could modulate this risk. Pregnant females are also susceptible to injury as a result of additional spinal loads due to the weight gain (~10-15 kg (Noon and Hoch, 2012; Schieve et al., 2000; Yaktine and Rasmussen, 2009)) although radical hormonal changes during pregnancy and associated biomechanical alterations are additional influential factors. According to our findings, slightly higher spinal loads in females (at identical BH, BW and age) combined likely with other risk factors (e.g., psychological factors, physical ability, job assignments (Bielby and Baron, 1986)) could play a role in greater prevalence of low back pain in females reported in some studies (DePalma et al., 2012; Schneider et al., 2006). Although current results suggest that age (in the range of 35-60 years) does not affect spinal loads, ageing could reduce damage tolerance threshold of intervertebral discs making them more susceptible to injury (Adams et al., 2015; Adams and Roughley, 2006).

In summary, using an image-based scaling algorithm for the trunk musculature and passive properties as well as prescribed (based on available *in vivo* measurements) spinal kinematics and lumbopelvic rhythm (based on sex and age) in conjunction with a novel technique of estimating trunk weight distribution in overweight and obese individuals, we investigated the effect of various personal factors (i.e., age, sex, BH and BW) on spine loads. Variations in BW have the greatest influence on spinal loads followed by those in sex, BH and age. With no load in hands, the rate of increase in spinal loads actually exceeds that in BW which highlights the exponential increase in spinal loads and hence risk of injury with BW especially in obese individuals. At identical BH and BW, spinal loads are slightly larger (~4.7% in compression and ~8.7% in shear) in females than in males.

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Table 1: Personalized factors and corresponding levels in the full factorial simulation design

Factors	Unit	Levels
Sex	-	Female – Male
Age	year	35 – 47.5 – 60
BH	cm	150 – 170 – 190
BW	kg	55 – 71.25 – 87.5 – 103.75 – 120

Table 2: Contribution (%) of each factor in all simulated tasks to the total sum of spinal loads squared.

Task	Response	Sex	Age	BH	BW
Standing with 5 kg load	L4-L5 Compression	0.2	0.3	0.4	99.2
	L5-S1 Compression	0.2	0.3	0.5	98.9
	L4-L5 Shear	13.6	5.4	32.0	49.0
	L5-S1 Shear	0.5	0.4	0.0	99.1
Flexion at 20º with no load	L4-L5 Compression	0.2	0.0	0.0	99.8
	L5-S1 Compression	0.6	0.0	0.1	99.2
	L4-L5 Shear	4.3	0.3	3.4	92.0
	L5-S1 Shear	1.1	0.1	0.5	98.3
Flexion at 50º with no load	L4-L5 Compression	0.1	0.0	0.1	99.7
	L5-S1 Compression	0.5	0.0	0.3	99.2
	L4-L5 Shear	4.3	0.4	4.4	91.0
	L5-S1 Shear	1.0	0.1	0.9	98.0
Flexion at 20º with 10 kg load	L4-L5 Compression	0.9	0.0	0.8	98.3
	L5-S1 Compression	1.5	0.0	1.0	97.5
	L4-L5 Shear	3.7	0.8	2.2	93.2
	L5-S1 Shear	2.4	0.6	1.5	95.5
Flexion at 50º with 5 kg load	L4-L5 Compression	0.7	0.1	0.9	98.3
	L5-S1 Compression	1.3	0.0	1.2	97.4
	L4-L5 Shear	3.7	0.9	2.6	92.8
	L5-S1 Shear	2.2	0.6	1.8	95.4
All tasks	L4-L5 Compression	0.5	0.0	0.3	99.2
	L5-S1 Compression	1.0	0.0	0.5	98.5
	L4-L5 Shear	4.1	0.6	3.2	92.2
	L5-S1 Shear	1.6	0.2	1.1	97.1
Overall	Compression	0.7	0.0	0.4	98.9
	Shear	2.1	0.3	1.5	96.1

Table 3: Effects of changing personal parameters (sex, age, BH and BW) on model parameters (Anderson et al.,2012; Ghezelbash et al., 2016; Shi et al., 2014)

Parameter	Passive Joint Properties	Muscle Moment Arms	Lumbopelvic Ratio	External Moment
Sex*	↘	↘	↘ or ↗	No Change
Age	↗	↗	↘	No Change
BH	↗✝	↗	No Change	↗
BW	↗	↗	No Change	↗

↗: increase; ↘: decrease; * here, sex is altered from male to female; ✝ increasing BH stiffens passive properties when BMI is kept constant.

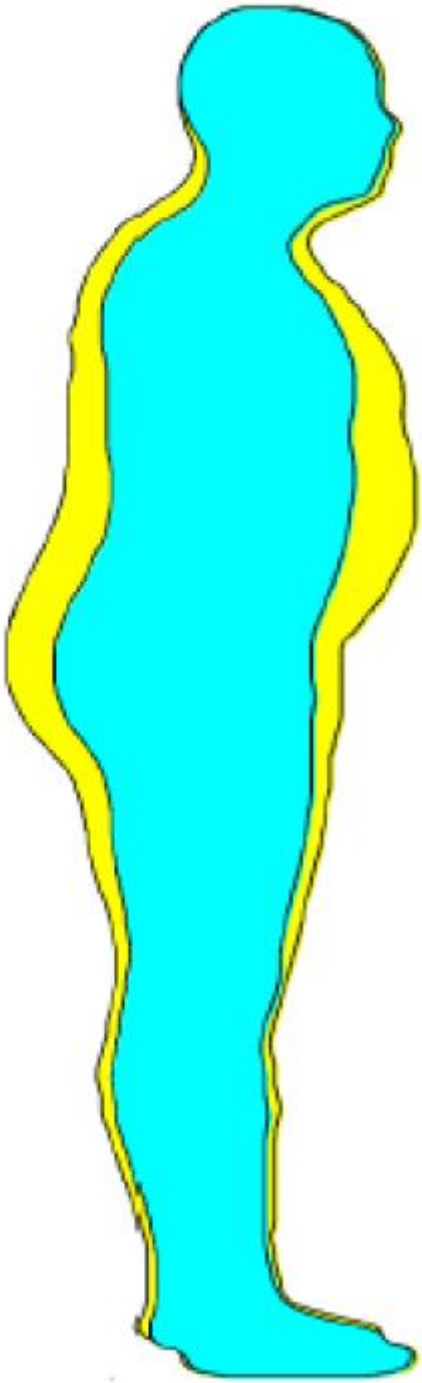

Fig. 1: Schematic body shape of an obese person (outer contour) versus a lean person (inner contour) in the sagittal plane (BMI Visualizer, Perceiving Systems Department, Max Planck Institute for Intelligent Systems, Germany).

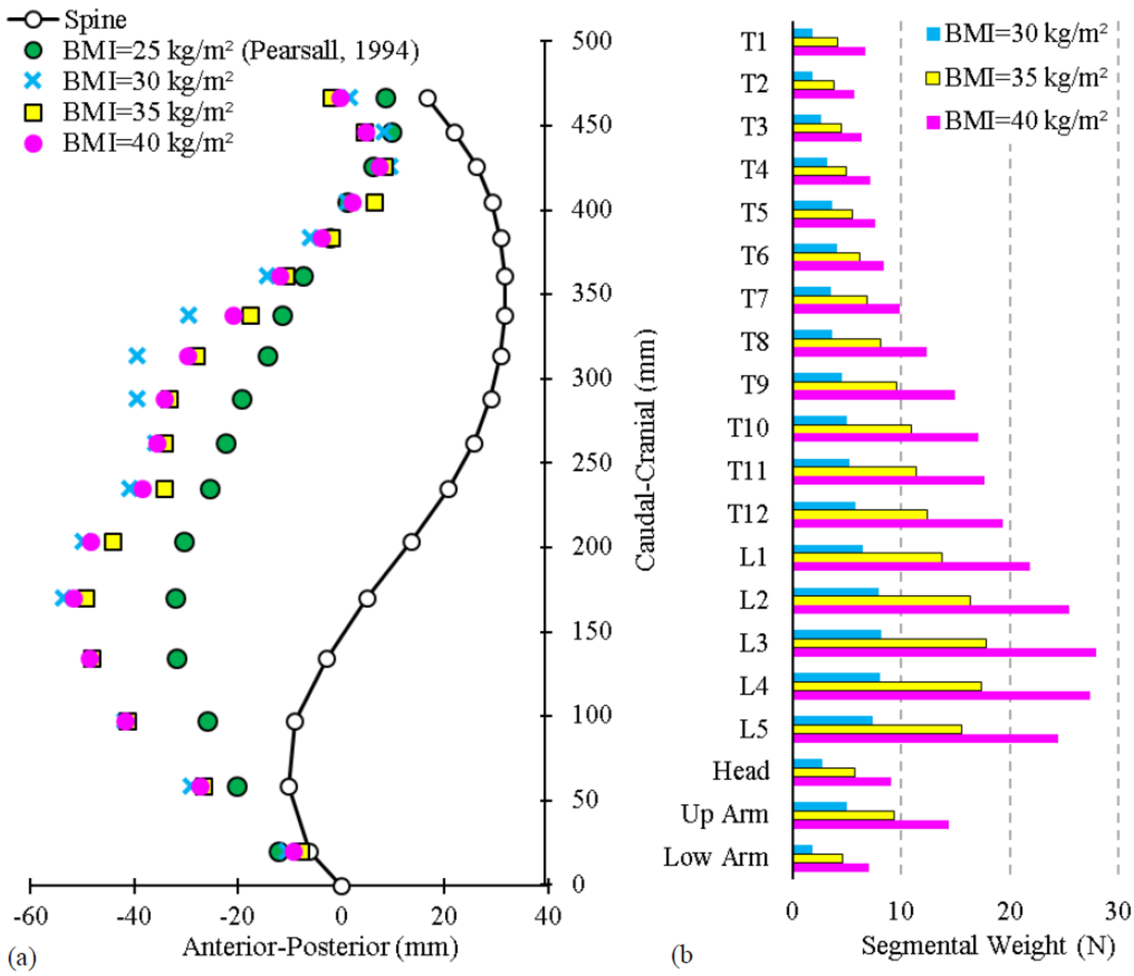

Fig. 2: Calculated (a) mass centers for BMI=25 kg/m^2 and for the additional trunk fat in cases with BMI>25 kg/m^2 and (b) additional (on top of those for BMI=25 kg/m^2) segmental weights of the trunk, arms (on each side and applied in the model onto the T3 level) and head (applied in the model onto the T1 level) for the overweight and obese cases with BH=173 cm.

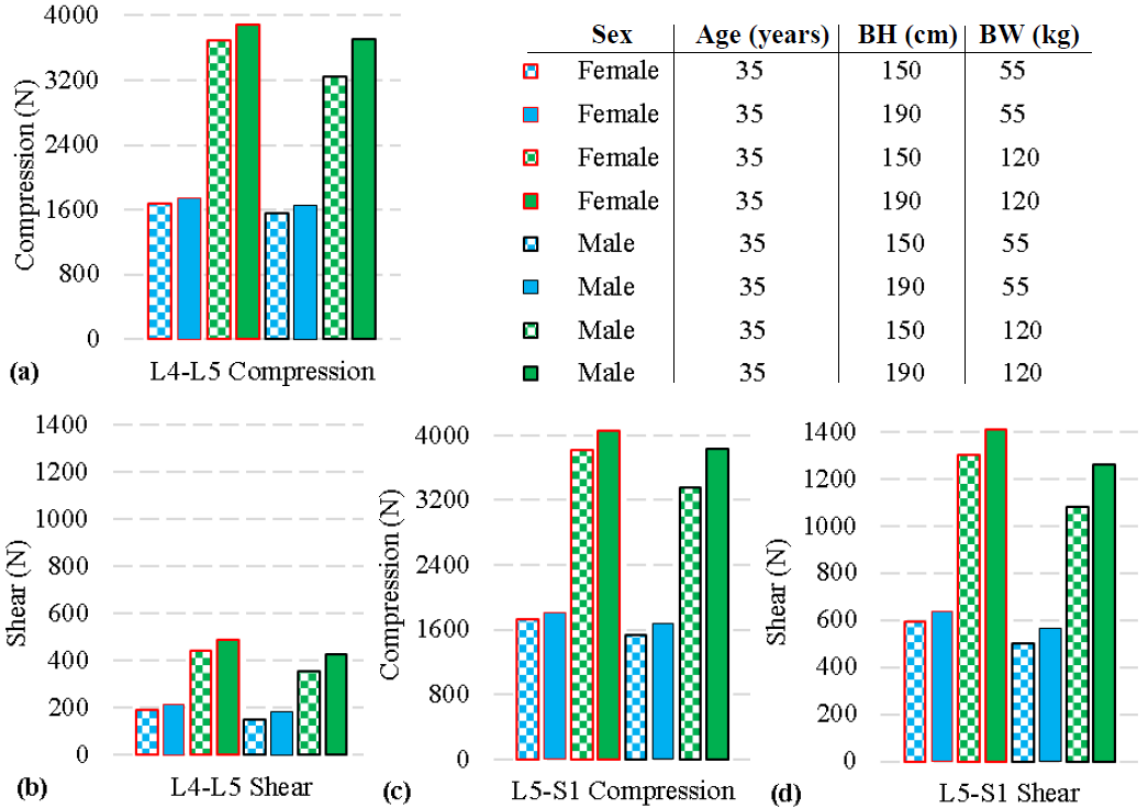

Fig. 3: Local shear and compression forces at the L4-L5 and L5-S1 levels with 5 kg load in hands at the trunk flexion of 50^{o} for various individuals of 35 years age.

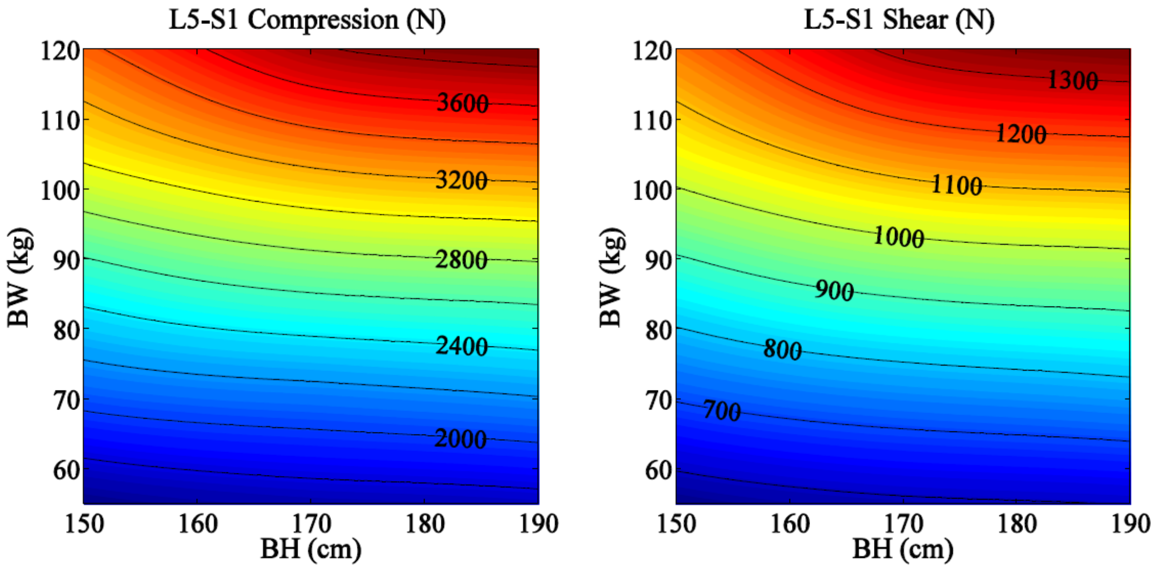

Fig. 4: **Local** compression (left) and shear (right) forces at the L5-S1 level at 50° flexion with 5 kg load in hands. Age and sex are set constant at 47.5 years and male.

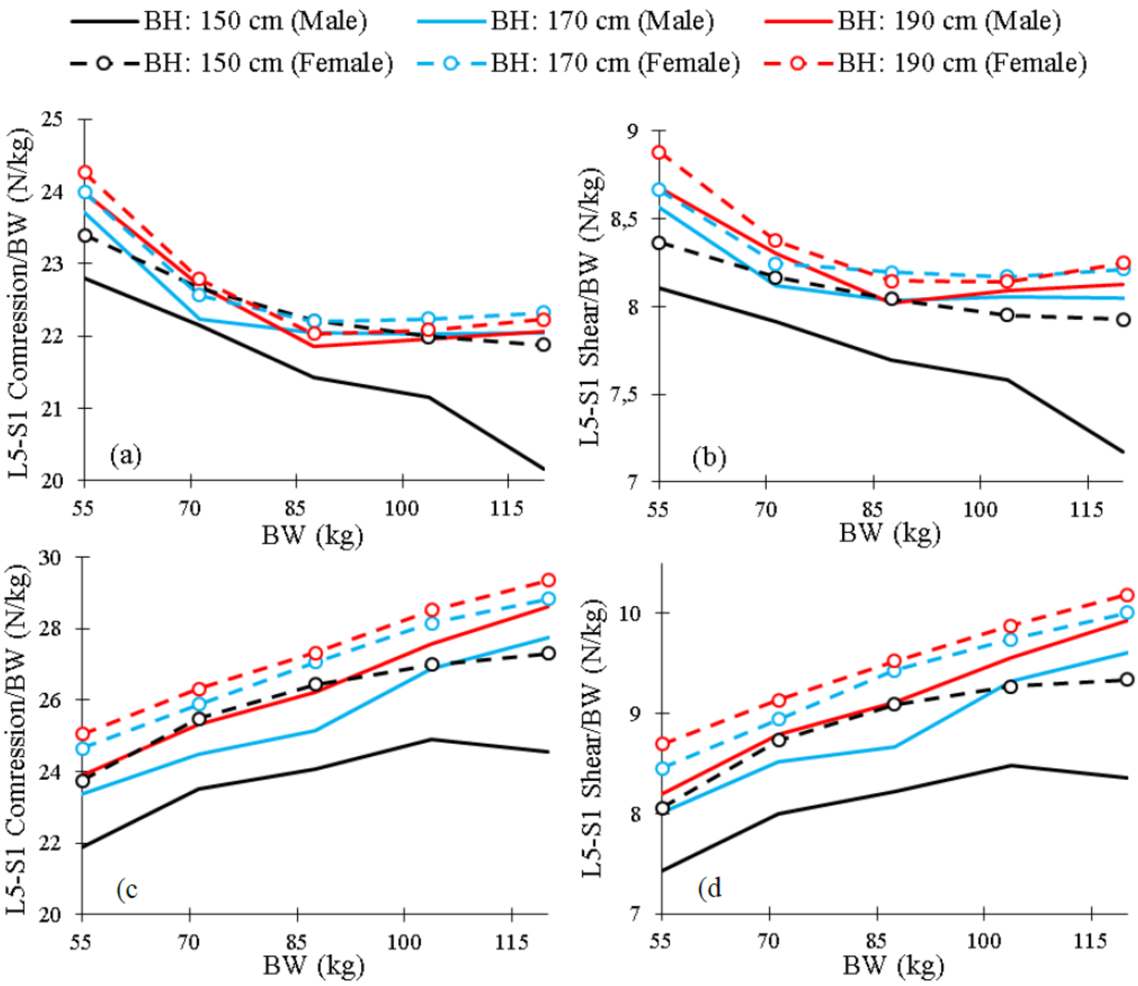

Fig. 5: BW-normalized local compression (left) and shear (right) forces at the L5-S1 level for 3 different BHs and 2 sexes under (a,b) 20° flexion with 10 kg load in hands and (c,d) 50° flexion without external load. Age is set at 47.5 years.

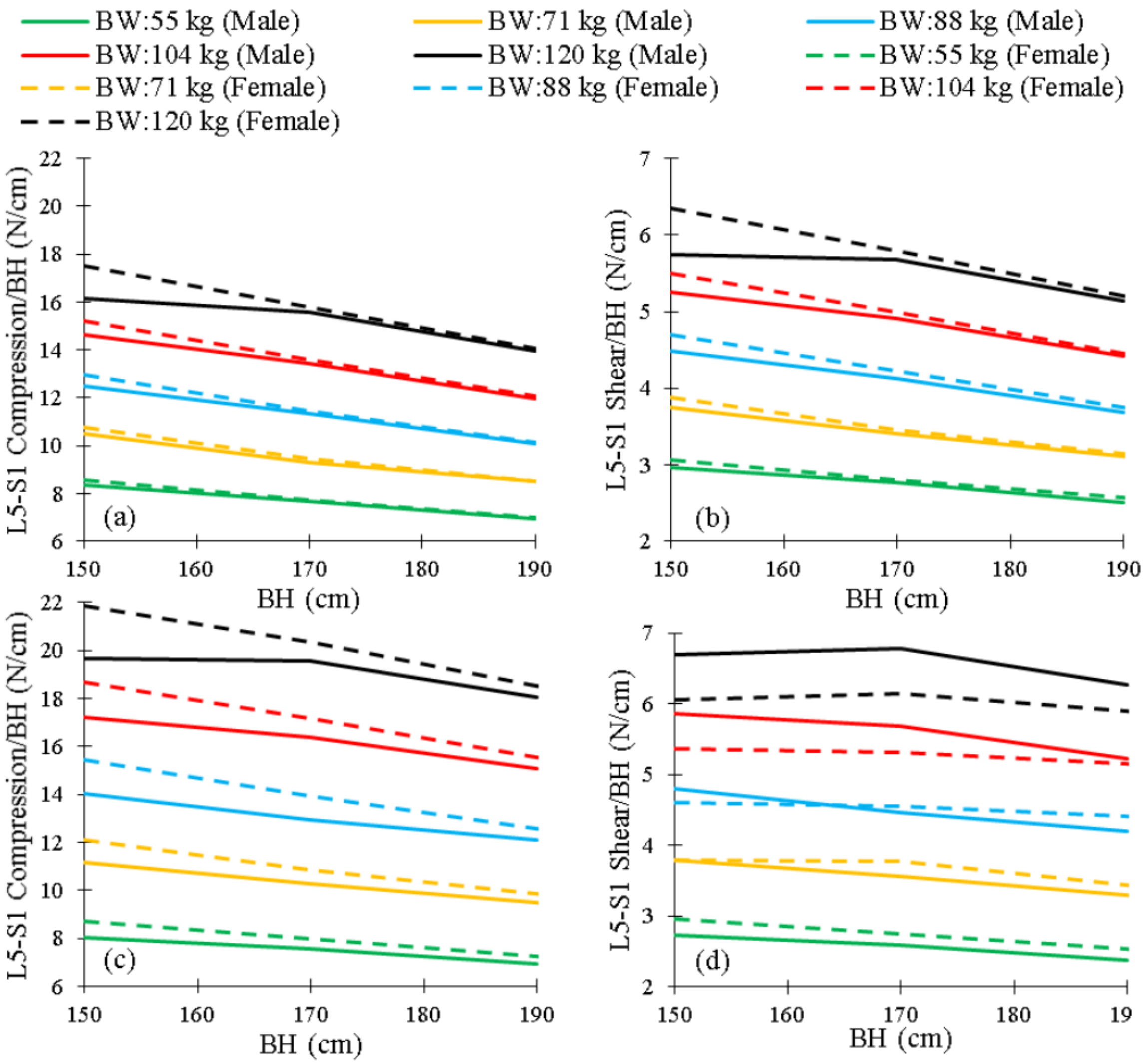

Fig. 6: BH-normalized local compression (left) and shear (right) forces at the L5-S1 level for 5 different BWs and 2 sexes under (a,b) 20° flexion with 10 kg load and (c,d) 50° flexion without load. Age is set at 47.5 years.

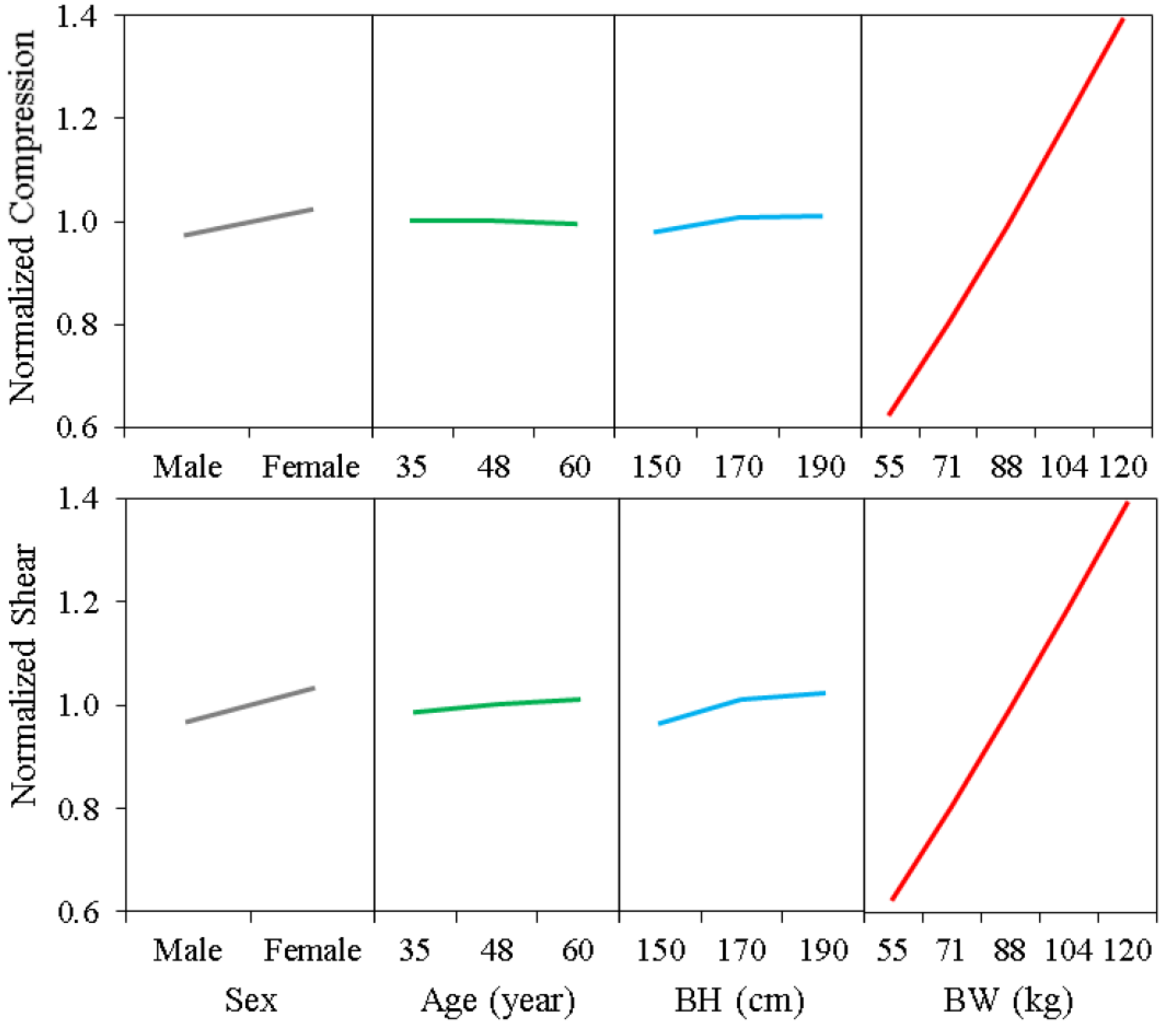

Fig. 7: Main effect plots of all simulations for compression (top) and shear (bottom) forces normalized to the mean values at each task at the L5-S1 disc.