The Mechanical Loading of the Spine in Physical Activities

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Cite as: Schäfer, R., Trompeter, K., Fett, Heinrich, K., Funken, J., Willwacher, S., Brüggemann, G.-P., Platen, P. (2022) The Mechanical Loading of the Spine in Physical Activities. *SportRxiv.*

Supplement: https://osf.io/vecgy/

Abstract

The purpose of this study was to summarize the mechanical loading of the spine in different activities of daily living and sports. Since the direct measurement is not feasible in sports activities, a mathematical model was adapted and applied to quantify spinal loading of more than 600 physical tasks in more than 200 subjects. The data demonstrate the excessive compression forces on the lumbar spine in some sport-related activities which are much higher than forces reported in normal daily activities and work tasks. Especially ballistic jumping and landing skills yield high estimated compression at L4/L5 of more than ten times body weight. Jumping, landing, heavy lifting and weight training in sports demonstrate compression forces significantly higher than guideline recommendations for working tasks. These results may help to identify acute and long term risks of low back pain and, thus, may guide the development of preventive interventions for low back pain or injury in athletes. Athletes and coaches could use the data in monitoring spinal demanding tasks and balance them out over the course of training periods.

Keywords: mechanical loading, lumbar spine, dynamic physical activities

Introduction

Mechanical loading of the spine during physical activity plays a significant role in the aetiology of back injuries and pain [1–6]. In-depth knowledge of the loads during different physical activities is mandatory for effective risk assessment, risk prevention, and possible modifications of spinal loads in everyday life up to athletic training and competition. For the latter, it can be assumed that the loads are substantially higher and occur more frequently in most sports disciplines.

The mechanical tissue loads that occur during highly dynamic movements in sports are discussed as a possible cause of back pain [7,8]. Current epidemiological studies also show the extreme prevalence of back pain in athletes and the clear differences in sports with different load profiles [9]. For the prevention of lifting-related low back pain among workers, NIOSH (National Institute for Occupational Safety and Health), recommends a limit of compression force at the lumbar spine of 3,400 N for men[10]. Assuming an average body mass for adult males of 85 kg, this results in a recommended load limit of 4 times body weight. This is assumed to be safe for regular loading[10]. The frequency, however, in that guideline is not restricted by the count of activities in a certain range, but by energy expenditure (2.2 – 4.7 kcal/min)[10]. While the NIOSH guideline may be applicable for the working context, they are not originally conceived for sports. While we assume higher values than 3400N regularly for athletes, knowing the loading and frequency of the activities allows us to judge the demands of the ´working´ athlete. This knowledge might be of importance for preventing low back pain and injury as coaches can try to balance the spine demanding tasks and progress spinal loading for – in particular adolescent – athletes to facilitate structural and functional adaptations.

However, examining the mechanical loading is not straightforward. Direct and usually invasive measurement methods –for example in Wilke et al.[11] - for quantifying the load on the spine cannot generally be used for load analysis in the highly dynamic movements of various sports. The risk of damage and infection associated with the installation of sensors in biological structures prohibits direct measurements. Also, such experiments are hardly ethically justifiable. Mathematical models can be used to calculate the torque and force in various parts of the body, which is a feasible, non-invasive method to estimate mechanical loading for highly dynamic tasks like in sports[12].

Nevertheless, few studies measured the mechanical loading of the spine in sports or physically demanding activities. For daily activities such as standing, sitting, lying, lifting and carrying, direct measurements of the intervertebral disc pressure and thus the resulting compression force at the L4/L5 motion segment are available [11,13,14]. For example, intradiscal measurements yield values of about 0.5 MPa for standing and 2.3 MPa for lifting [11]. Lifting loads of 150 kg yielded compression forces higher than 9,500 N[15]. Powerlifters were shown to reach lumbar compression forces higher than 15,000 N lifting 285kg [16].

This study aims to estimate the mechanical loading of the spine for the most common activities of daily life and among many sports disciplines. These data can be used as a solid reference in further research or to predict spinal loading by practitioners. As an adjunct aim, we want to verify the estimated values by comparing them with the few available values from the literature and highlight the potential for further usage.

Methods

Participants

To quantify the mechanical load occurring in activities of sports and daily life, we summarized the data from several investigations between 2011 and 2018 conducted within a larger study project. For these investigations, we recruited female and male athletes aged 16 to 32 from 16 Olympic sports disciplines. All athletes were competing at a high competitive performance level All participants gave their written informed consent before the study began. The study was conducted in agreement with the Declaration of Helsinki and approved by the medical ethics committee of the Ruhr University Bochum (Reg.-Nr.: 4904-14).

Study Design

The participants performed several sport-specific movements in their respective disciplines and common daily activities in our experimental setup. Thus, the various activities were carried out by different samples since a large number of the activities could only be performed by athletes of the respective sports disciplines in a repeatable manner and at a suitable technical level. Each movement was performed as close to competition as possible and was repeated three times. The attempt that the athletes themselves felt was the most realistic was then taken for further analysis.

We selected the most frequent activities in the respective disciplines for investigation. This selection process was based on competition observations and consensus with elite athletes and coaches. Collisions, opponent interactions, support or disturbance of movements and any traumatic events with spontaneous tissue failure were not considered.

We chose the mathematical modelling approach to quantify the mechanical loading of the lumbar spine, since direct measurements on spinal motion segments were ruled out for technical and, above all, ethical reasons.

Measurement

Kinematic data were generated using 3D motion analysis (VICON Nexus, 12 MX40 cameras, 200 Hz, recursive Butterworth filter with 12 Hz cut-off frequency). For this purpose, 54 retroreflective markers were applied to the lower and upper limb segments, trunk, and spine of the subjects. Ground reaction forces were recorded using force platforms embedded in the ground. Force data were recorded synchronously with motion data and sampled at 1,000 Hz.

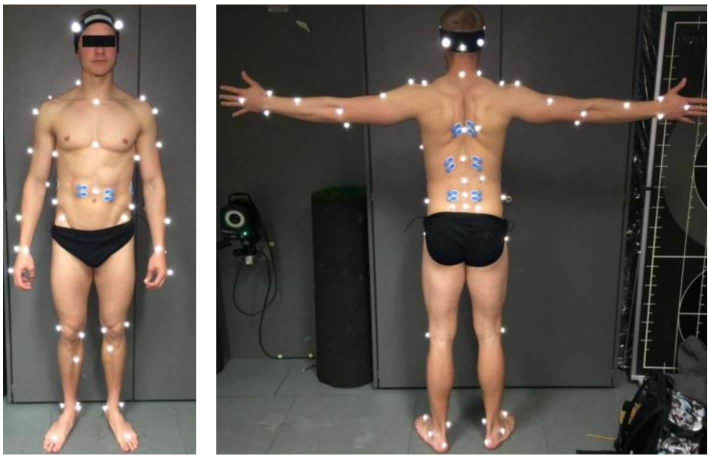

Figure 1 | Experimental setup of the 54 markers.

Muscle activity of the right and left erector spinae, latissimus dorsi, psoas, obliquus externus, and rectus abdominis muscles were recorded by 8-channel surface electromyography (EMG) with a sampling rate of 2,000 Hz per channel and telemetrically registered (Myon AG, Schwarzenberg, Switzerland) and recorded synchronously with the movement and reaction force data. For EMG application, after hair removal and skin preparation, the areas of the muscle bellies were cleaned with alcohol and Ag/AgCl surface electrodes (sensor area: 15 mm, Ambu Blue Sensor N, Ambu A/S, Ballerup, Denmark) were applied with conductive gel parallel to the muscle fibres of the muscles under investigation with an electrode spacing of about 2.3 cm.

The mathematical model

The external torques and forces at L4/L5 required for the mathematical model are implemented by inverse-dynamic modelling with a multibody model [17]. This model assumes force and torque to be distributed around the biological structures like intervertebral discs, vertebral bodies, ligaments and muscles. However, the distribution quantities remain unknown, and the number of unknown variables typically exceeds the number of equations available to describe the system mechanics. To reduce the number of unknown variables, non-trivial assumptions are necessary. We use mathematical optimization methods incorporating physiological data, such as muscle activity (EMG) and setting physiological boundaries for the parameters in the model.

Torque and Force

The force-transmitting structures considered in this model are muscles, ligaments and vertebral bodies (including the intervertebral discs). This results in the equations:

$$F = \sum f_i^m + \sum f_i^l + \sum f_i^c$$

$$M = \sum f_i^m r_i^m \sum f_i^l r_i^l \sum f_i^c r_i^c$$

The intersegmental forces and torques of the joint centre are given by F and M, respectively. The *f*-vectors represent the forces transmitted via the muscles (f^m), the ligaments (f^l) and the vertebral body including the intervertebral disc (f^c). The radii r^m, r^l und r^c are the respective lever arms at the joint centre. In the distribution problem, F and M are assumed to be known and inversely dynamically calculated. The forces f^m, f^l und f^c are going to be calculated.

Muscles

To reduce the complexity of the model, we accepted 4 muscle groups as major force-transmitting structures: M. rectus abdominis, M. obliquus externus and internus, M. erector spinae and M. latissimus dorsi. Thus, smaller muscle groups were neglected due to their cross-sectional volume and probably minor contribution to the generated force.

The lever arms, the maximal, the muscle cross-sectional area for estimating the maximum force, and the muscle pull direction were estimated based on MRI images from a small sample of athletes in our study. In some cases, additional information was taken from the literature and scaled based on the anthropometric data of the subjects.

Ligaments

The ligaments as force-transmitting structures were omitted for further simplification since their task is to guide the joint. Moreover, the morphological situation could only be recorded extremely imprecisely from magnetic resonance imaging. Furthermore, there is less information available in the literature on the mechanical properties of the ligaments of the spine, especially for younger people.

Vertebral Body

The force at the vertebral body was our parameter of interest. In our model, we further omitted the facet joints due to the relatively small contact area. The sum of the mechanical loads on this joint surface generates the resulting contact force and moment. The resulting contact force can in turn be decomposed into a compression force and a shear force component.

Electromyography (EMG)

To further reduce the number of unknown variables, we used EMG measurements of the muscles included in the model. Individual muscles were identified as inactive for certain time intervals and thus switched off in the model and to not transmit force anymore. In case the measurement was not possible due to technical or pragmatic reasons, agonistic muscles were considered active and antagonistic muscles switched off all the time. Therefore our model-based spinal load calculations are conservative and represent the minimum of the real load to the spinal structure

Mathematical optimization

In a final step, mathematical optimization using a cost function was used to find a solution for the equation system. This assumes that the muscle forces for a given activity are selected and used according to the criterion of optimal functionality. In this study, after testing different cost functions, the square of the sum of the mechanical strain of the muscles involved (i.e., the square of the force related to the physiological cross-section) was chosen as the minimization criterion. Furthermore, the boundary conditions for the optimization were set as follows:

$$0 \leq f_i^m \leq a_i^m \text{ und } 0 \leq f_i^c \leq a_i^c$$

Due to morphological and physiological constraints, a_i^m and a_i^c are the maximum possible forces that can occur in the muscle and the join, respectively. Furthermore, muscle and contact forces cannot be negative.

Data Processing
The variables in this dataset regarding the segment L4/L5 are:

- Torque [Nm] ($f^c r^c$)
- Relative Torque in [Nm/kg] ($f^c r^c$ / *body weight* [kg])
- Compression [N] (f^c)
- Compression relative to body weight [AU] (f^c / *body weight* [N])
- Time [s or ms]
- “Spinal load” [AU]: Relative compression integrated over time

The lever arm r^c was set to 5 cm. Bodyweight in N was calculated by multiplying with the gravitational force (g = 9.81 m/s²). The parameters were determined in an interval at >80% of the maximum of the compression force during the respective movement or posture.

Statistics
Descriptive values – mean (standard deviation) - are presented for each activity categorized by the tested population. Further, bootstrapped non-parametric 95 % confidence intervals – $CI_{95\%}$ [lower limit; upper limit] - are calculated within each factor and each parameter. For certain activities (running, lifting, jumping, standing), additional simple linear regression, multilevel and spline models were fitted and compared. Due to the explorative nature of this analysis, no formal significance test was applied. All computations were done in R v4.0.4[18]. The full package list and the details of our analyses can be viewed in our reproducible R-Markdown script (see appendix).

Results

Our dataset contains 637 observations from 248 participants. After filtering activities with only one observation and overhead activities, we got 578 observations from 236 participants, 18 groups (e.g., basketball, volleyball, hockey) and 67 investigated activities. Participant characteristics are shown in Table 1.

Table 1 | Population Characteristics

population	n	Body Mass [kg]	Height [cm]	Sex f/m
basketball	10	82.3 (7.6)	187.4 (5.4)	1/9
bob	8	83.9 (10.1)	180.5 (8.1)	4/4

get_up	9	78.8 (3.7)	178.7 (5.2)	0/9
hockey	12	70.9 (10.8)	170.8 (8.9)	7/5
hurdle	3	72 (5.3)	179.3 (10.2)	2/1
javelin	20	76.1 (11.4)	182.2 (9)	8/12
jumper	13	78.2 (4.2)	189.7 (6.3)	0/13
lifter_amateur	31	71.2 (5)	173.6 (5.9)	13/18
lifter_elite	17	80.5 (8.1)	178.5 (6.2)	0/17
pick_up	10	65.8 (5.8)	170.8 (4.3)	10/0
rowing	10	79.7 (5.6)	181.7 (6.1)	0/10
runner	20	69 (8.7)	173.3 (6.5)	0/20
shotput	13	74.2 (12.8)	177.3 (11.9)	6/7
standing	15	73.4 (10.8)	181.4 (8.7)	6/9
students	20	72.7 (8.9)	175.2 (9.6)	10/10
tennis	4	73.2 (10.5)	175.8 (4.8)	2/2
volleyball	11	77.3 (7.3)	181.5 (9.1)	6/5
walk	10	71.1 (9.4)	174.1 (6.1)	4/6
Total	**236**	**74.5 (9.4)**	**178 (8.9)**	**79/157**

The results of the studied activities are summarized in Table 2.

Table 2 | Outcomes by population

Population / Activity	n	Compression [N]		Normalized Compression [AU]		Time [ms]	
		mean (SD)	$CI_{95\%boot}$	mean (SD)	$CI_{95\%boot}$	mean (SD)	$CI_{95\%boot}$
standing							
standing 0deg	15	596 (184)	[508; 685]	0.9 (0.3)	[0.7; 1]	749 (348)	[575; 911]
standing 30deg	15	1271 (351)	[1094; 1436]	1.8 (0.6)	[1.5; 2.1]	1249 (348)	[1061; 1411]
standing 90deg	15	2195 (560)	[1921; 2461]	3.1 (0.9)	[2.7; 3.5]	1549 (348)	[1371; 1705]
walk							
walk	10	966 (171)	[858; 1060]	1.4 (0.2)	[1.3; 1.5]	153 (47)	[129; 182]
uphill stairs	10	1206 (212)	[1072; 1326]	1.7 (0.4)	[1.6; 2]	230 (70)	[192; 276]
get up							
sit stand	9	2384 (622)	[1986; 2780]	3.1 (0.9)	[2.6; 3.7]	712 (189)	[601; 826]
pick up							
knees flexed	10	271 (208)	[168; 403]	0.4 (0.3)	[0.3; 0.6]	178 (62)	[146; 220]
knees straight	10	2170 (156)	[2083; 2268]	3.4 (0.3)	[3.2; 3.6]	311 (85)	[263; 359]
basketball							
layup	3	565 (284)	[258; 820]	0.7 (0.3)	[0.4; 1]	19 (10)	[8; 28]
jumpshot takeoff	2	646 (283)	[446; 846]	0.9 (0.4)	[0.5; 1.2]	72 (96)	[4; 140]
pass	3	769 (645)	[240; 1488]	1 (0.8)	[0.3; 1.9]	160 (112)	[44; 268]
shot	3	722 (490)	[210; 1186]	1 (0.7)	[0.3; 1.6]	57 (58)	[16; 124]
pass overhead	4	800 (59)	[745; 850]	1.1 (0.2)	[1; 1.2]	33 (15)	[19; 44]
jumpshot	3	835 (600)	[402; 1520]	1.1 (0.8)	[0.6; 2.1]	21 (20)	[8; 44]
catch pass	3	874 (337)	[546; 1220]	1.1 (0.5)	[0.7; 1.7]	80 (94)	[16; 188]
pass onehand	2	970 (467)	[640; 1300]	1.4 (0.8)	[0.8; 1.9]	122 (122)	[36; 208]
takeoff	2	1278 (25)	[1260; 1296]	1.6 (0)	[1.6; 1.6]	48 (51)	[12; 84]
sidestep	4	1263 (974)	[513; 2057]	1.7 (1.3)	[0.7; 2.7]	143 (58)	[89; 186]
rebound jump	2	1467 (1252)	[582; 2352]	2.1 (1.7)	[0.9; 3.2]	80 (11)	[72; 88]
layup takeoff	3	2167 (1114)	[1080; 3306]	3 (1.4)	[1.5; 4.3]	40 (17)	[28; 60]
powermove jump	3	2499 (1385)	[1678; 4098]	3.4 (1.7)	[2.3; 4.4]	55 (47)	[4; 96]

Population / Activity	n	Compression [N] mean (SD)	Compression [N] $CI_{95\%boot}$	Normalized Compression [AU] mean (SD)	Normalized Compression [AU] $CI_{95\%boot}$	Time [ms] mean (SD)	Time [ms] $CI_{95\%boot}$
dribbling	8	4624 (742)	[4173; 5122]	5.5 (0.6)	[5.1; 5.9]	129 (27)	[112; 147]
bob							
start	8	8976 (2120)	[7662; 10333]	10.8 (1.4)	[9.8; 11.6]	*NA*	*NA*
hockey							
running moderate	9	1138 (318)	[954; 1324]	1.6 (0.3)	[1.4; 1.8]	154 (50)	[125; 185]
block shot	5	1056 (458)	[704; 1410]	1.7 (0.8)	[1.1; 2.3]	296 (143)	[184; 405]
scoop	5	1110 (494)	[760; 1460]	1.7 (0.7)	[1.3; 2.3]	283 (56)	[235; 324]
hit	5	1126 (245)	[941; 1315]	1.8 (0.4)	[1.5; 2.2]	165 (108)	[80; 243]
push	5	1199 (582)	[739; 1657]	1.8 (0.8)	[1.3; 2.4]	146 (106)	[56; 226]
running fast	9	2622 (671)	[2229; 3023]	3.6 (0.6)	[3.3; 4]	136 (36)	[115; 159]
dribbling	8	2546 (1088)	[1817; 3211]	3.6 (1.4)	[2.8; 4.5]	272 (59)	[235; 311]
running cod	7	3166 (1246)	[2303; 4037]	4.8 (2.1)	[3.4; 6.3]	171 (89)	[116; 235]
arg backhand	7	3457 (760)	[2923; 3947]	5 (1.2)	[4.1; 5.8]	89 (48)	[56; 119]
hurdle							
land	3	1933 (102)	[1834; 2038]	2.7 (0.1)	[2.7; 2.9]	30 (7)	[25; 38]
cross	3	1996 (105)	[1888; 2098]	2.8 (0.1)	[2.7; 3]	30 (11)	[17; 38]
jump	3	3450 (182)	[3274; 3638]	4.9 (0.2)	[4.8; 5.2]	19 (5)	[15; 25]
javelin							
javelin throw	7	316 (129)	[224; 407]	0.4 (0.2)	[0.3; 0.6]	142 (56)	[108; 181]
prep	13	413 (155)	[339; 500]	0.6 (0.2)	[0.5; 0.6]	135 (23)	[123; 148]
jumper							
high jump	8	6631 (841)	[6095; 7142]	8.8 (1)	[8.2; 9.5]	88 (15)	[79; 98]
long jump	5	12456 (1359)	[11468; 13608]	15.7 (1.5)	[14.5; 16.8]	69 (10)	[62; 78]
lifter amateur							
lateral lift 2x20kg dumbell	11	1826 (346)	[1641; 2045]	2.5 (0.4)	[2.3; 2.7]	591 (119)	[523; 657]
lift 10kg barbell	20	2632 (479)	[2432; 2841]	3.9 (0.7)	[3.6; 4.2]	324 (131)	[271; 380]
lifter elite							

Population / Activity	n	Compression [N] mean (SD)	Compression [N] $CI_{95\%boot}$	Normalized Compression [AU] mean (SD)	Normalized Compression [AU] $CI_{95\%boot}$	Time [ms] mean (SD)	Time [ms] $CI_{95\%boot}$
clean 10kg	11	2211 (358)	[2003; 2384]	3 (0.4)	[2.7; 3.2]	591 (119)	[527; 656]
clean 20kg	11	2867 (367)	[2644; 3047]	3.8 (0.3)	[3.6; 4]	591 (119)	[523; 656]
clean 50kg	11	4782 (772)	[4405; 5236]	6.4 (0.8)	[6; 6.9]	600 (121)	[530; 669]
clean 80kg	6	7979 (443)	[7652; 8281]	9.2 (0.5)	[8.8; 9.5]	540 (122)	[449; 631]
clean jerk 100kg	3	8457 (1218)	[7098; 9452]	9.4 (1.1)	[8.2; 10.1]	610 (92)	[505; 675]
rowing							
stroke	10	5051 (372)	[4843; 5268]	6.5 (0.2)	[6.4; 6.6]	249 (61)	[215; 285]
runner							
running 2.5	20	3184 (670)	[2893; 3474]	4.7 (0.6)	[4.4; 5]	121 (26)	[111; 132]
running 3.5	20	4775 (1006)	[4354; 5225]	7 (1)	[6.6; 7.4]	109 (23)	[100; 119]
running 4.5	20	5681 (1167)	[5180; 6196]	8.4 (1.1)	[7.9; 8.9]	104 (22)	[95; 114]
running 5.5	20	6079 (1248)	[5544; 6668]	8.9 (1.2)	[8.4; 9.5]	93 (20)	[85; 102]
running 6.5	20	6988 (1435)	[6393; 7621]	10.3 (1.4)	[9.7; 10.8]	82 (17)	[75; 90]
shotput							
power toss	8	351 (106)	[284; 426]	0.5 (0.2)	[0.4; 0.6]	154 (99)	[90; 219]
shotput	8	1976 (1038)	[1320; 2626]	2.8 (1.5)	[1.8; 3.8]	135 (91)	[76; 196]
slide	20	2796 (926)	[2406; 3226]	3.8 (0.9)	[3.4; 4.2]	585 (289)	[460; 707]
power toss prep	8	3574 (1523)	[2593; 4535]	4.7 (1.9)	[3.3; 5.9]	278 (120)	[220; 365]
students							
cmj	20	3344 (507)	[3113; 3552]	4.7 (0.5)	[4.5; 4.9]	32 (15)	[26; 39]
dj 20	20	8359 (1267)	[7792; 8915]	11.7 (1.2)	[11.3; 12.2]	16 (8)	[13; 20]
dj 40	20	9613 (1457)	[8996; 10199]	13.5 (1.3)	[12.9; 14]	18 (8)	[14; 22]
dj 60	20	11953 (1812)	[11129; 12674]	16.8 (1.7)	[16.1; 17.5]	21 (10)	[17; 26]
tennis							
topspin	4	1196 (820)	[494; 1899]	1.6 (0.9)	[0.8; 2.4]	18 (16)	[4; 32]
volleyball							
lower pass lateral	2	1310 (651)	[850; 1770]	1.8 (0.9)	[1.2; 2.5]	116 (85)	[56; 176]
lower pass frontal	2	1433 (757)	[898; 1968]	1.8 (0.9)	[1.2; 2.5]	116 (0)	[116; 116]

Population / Activity	n	Compression [N]		Normalized Compression [AU]		Time [ms]	
		mean (SD)	$CI_{95\%boot}$	mean (SD)	$CI_{95\%boot}$	mean (SD)	$CI_{95\%boot}$
lower pass low	2	2000 (141)	[1900; 2100]	2.4 (0)	[2.3; 2.4]	62 (48)	[28; 96]
ready	8	2178 (758)	[1658; 2696]	2.9 (0.9)	[2.4; 3.5]	1706 (361)	[1462; 1919]
dig	2	4045 (2041)	[2602; 5488]	5.2 (2.4)	[3.4; 6.9]	20 (3)	[18; 22]
dive	2	4494 (1109)	[3710; 5278]	5.8 (1.2)	[4.9; 6.6]	40 (8)	[35; 46]

Mean: arithmetical mean, SD: standard deviation, CI95%boot: bootstrapped confidence interval using the smean.cl.boot() function from the Hmisc package, Normalized compression is derived by dividing compression by weight in N

Standing

When standing upright, the average estimated compression force at L4/L5 was 596 N ($CI_{95\%boot}$[508; 685]), which equals 0.93 ($CI_{95\%boot}$ [0.7; 1]) times the body weight. When leaning forward, a simple linear regression model yields an increase of absolutely 17.4 N ($CI_{95\%}$[14.2;20.6]) or relative to body-weight of 0.025 ($CI_{95\%}$[0.02;0.03]) per degree (see Figure 2A). The predictions from this model align well with observed values for normalized compression at 30° (1.8 times bodyweight, $CI_{95\%}$ [1.5, 2.1]) and 90° (3.1 times bodyweight, $CI_{95\%}$ [2.7; 3.5]) trunk reclination. Though, the variability increases on higher angles.

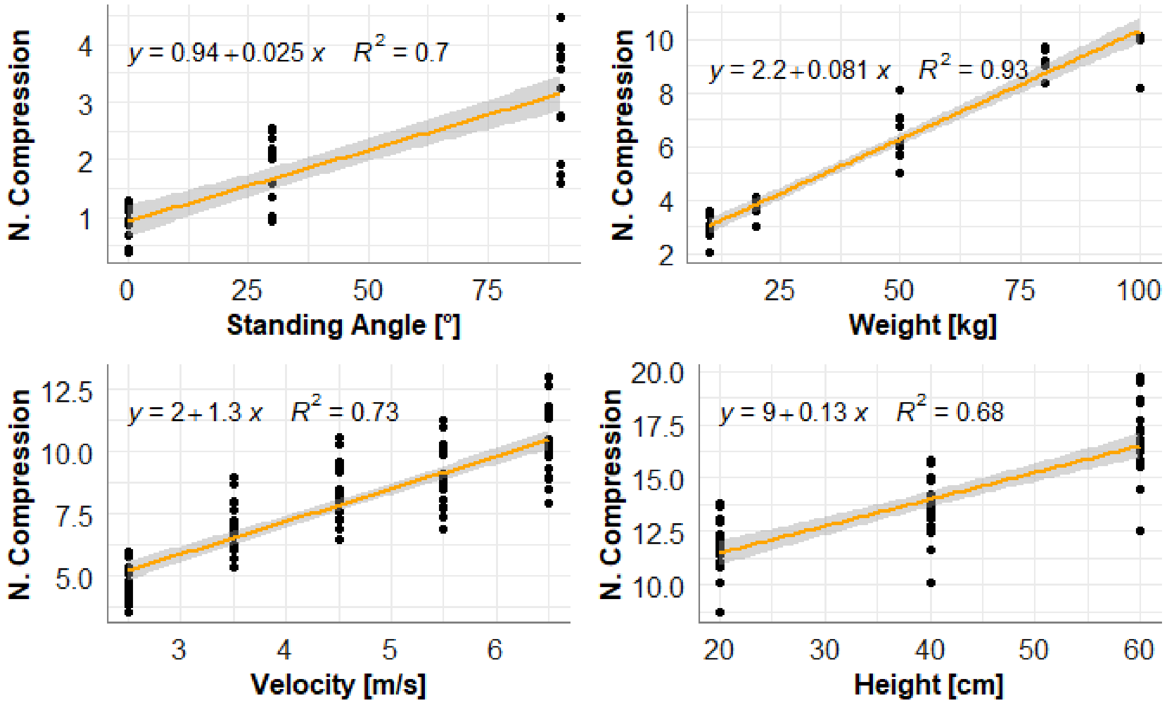

Figure 2 | Linear regression models for standing, lifting, running and jumping. Outcome (y-axis) is the normalized compression (compression [N] / bodyweight [N]). A: Standing with trunk inclination, model: compression ~ trunk angle [°], B: Cleans with barbell, model: compression ~ lifted weight [kg], C:constant running with different speeds, model: compression ~ running speed [m/s], D: dropjumps from different heights, model: compression ~ drop height [cm]

Lifting

When lifting a barbell of 10kg, the average estimated compression force at L4/L5 is 2632 N ($CI_{95\%}$ [2442;2861]), which equals 3.9 ($CI_{95\%}$ [3.6; 4.2]) times the body weight. The influence of lifting technique is striking: In a cohort of more experienced lifters, the estimated average force in lifting 10kg is considerably lower (normalized compression: 3.0 $CI_{95\%}$ [2.7; 3.2]). Also, lifting 2x20 kg dumbbells laterally resulted in even lower normalized compression force (2.5 $CI_{95\%}$ [2.2; 2,7]). The highest compression forces in this category were observed when pushing a bobsleigh from professional athletes: 8976 N ($CI_{95\%}$ [7596.8; 10264.4]), 10.8 ($CI_{95\%}$ [9.8; 11.7]) times body weight.
In a linear regression model, we evaluated the impact of weight lifted in cleans on compression. While the intercept predicts an average compression of 1352 N ($CI_{95\%}$ [1018; 1695]) when performing cleans with no weight, the increase per kg is 75 N ($CI_{95\%}$ [68; 82]). – see Figure 2b.

Walking/Running

When walking, the average estimated compression force at L4/L5 is 966 N ($CI_{95\%}$ [868; 1062]), which equals 1.4 ($CI_{95\%}$ [1.3; 1.5]) times the body weight. When jogging or running, the compression force increases to 1.6 ($CI_{95\%}$ [1.4; 1.8] and 3.6 ($CI_{95\%}$ [3.3; 4.0]) times the body weight, respectively.
In 20 runners, velocities from 2.5 m/s to 6.5 m/s with 1 m/s increments were tested. A linear regression model yields at an intercept of 2 m/s a compression force of 3113 N ($CI_{95\%}$ [2648; 3578]) or 4.6 ($CI_{95\%}$ [4.1; 5]). The predicted increase per 1 m/s in normalized compression is 1.3 ($CI_{95\%}$ [1.2; 1.5]). The linear model aligns well with velocities from 3.5 to 6.5, though there might be some non-linearity when increasing speed from 2.5 m/s to 3.5 m/s. Moreover, the time spent at these forces decreases over time, whereas the load (integral over time) increases up to 4.5 m/s and stays nearly the same until 6.5 m/s. The time intervals are relatively short (<100ms) but – as in the nature of running – highly repetitive.

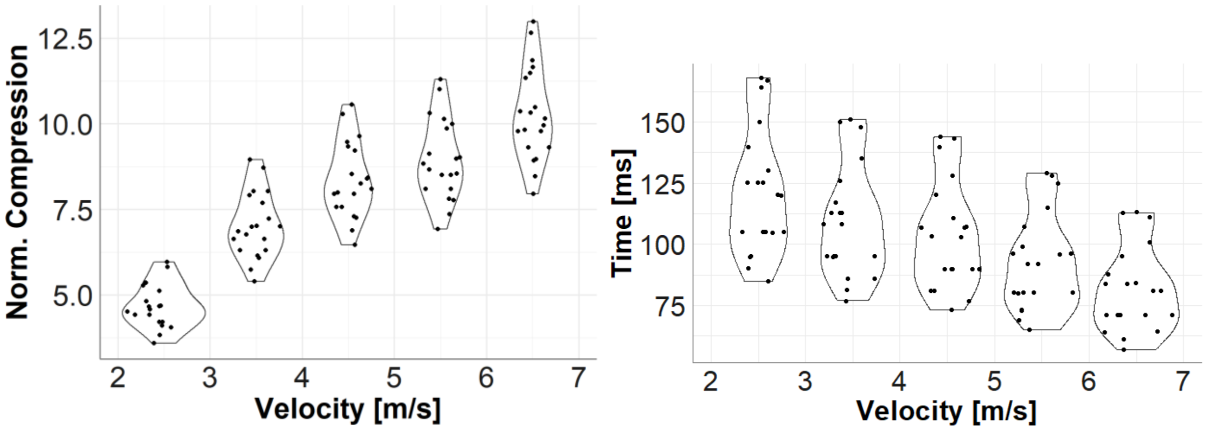

Figure 3 | Violin plots for running outcomes: normalized compression and time. A: increasing trend for compression with running speed. B: decreasing trend for time with running speed

Jumping

When performing a countermovement jump, the average estimated compression force at L4/L5 is 3343 N ($CI_{95\%}$ [3117; 3562]), which equals 4.7 ($CI_{95\%}$ [4.5; 4.9]) times the body weight (n=20). In professional high and long jumpers, the compression force increases to 8.8 ($CI_{95\%}$ [8.3; 9.5] and 15.7 ($CI_{95\%}$ [14.5;16.8]) times the body weight, respectively.
In 20 sports students, drop jump heights of 20cm, 40cm and 60cm were tested. A linear regression model predicts for 20cm drop-jump height an absolute compression force of 8178 N ($CI_{95\%}$ [7550; 8806]) and relative to bodyweight 11.5 ($CI_{95\%}$ [10.9; 12.1]) - see Figure 2D. The predicted increase per cm in normalized compression is 0.13 ($CI_{95\%}$ [0.10; 0.15]). The linear model fits the data quite well (R^2 = 0.68), but we would rather expect a curvilinear form and we also see increasing variability with increasing drop-jump height.

Figure 4 illustrates the relationship between movement speed or initial energy and compression forces at L4/L5 among different activities of daily living and sports.

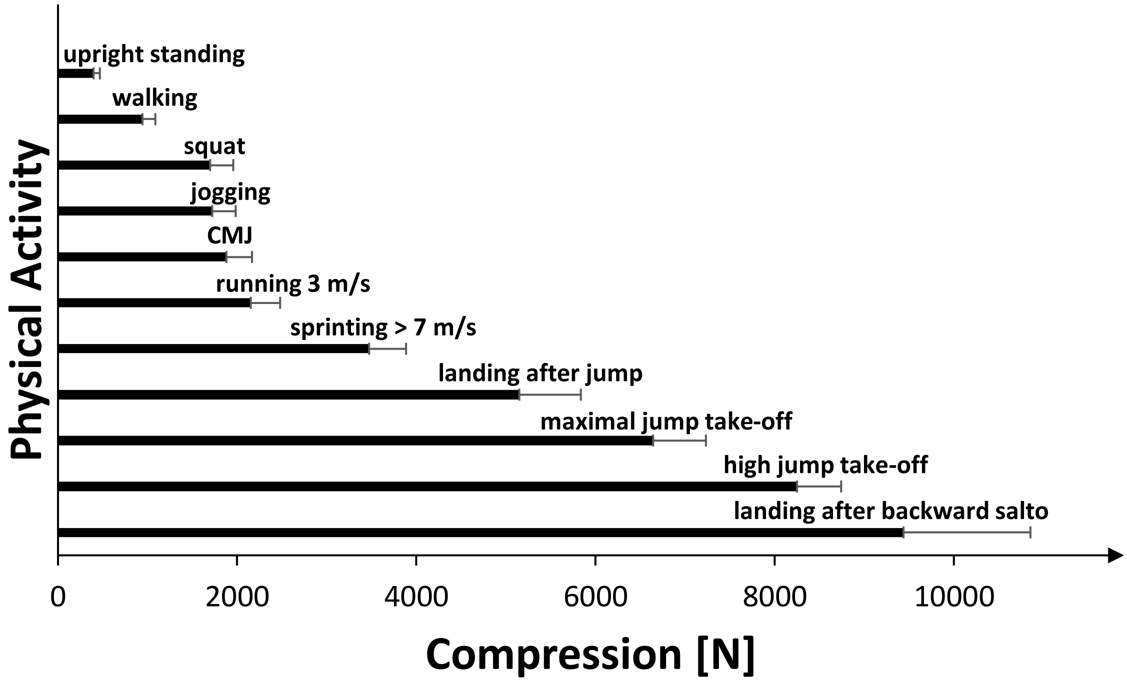

Figure 4 | Maximum compression forces at L4/L5 during different physical activities. All activities were studied under training conditions in the laboratory. It is to be expected that the loads in the competition situation and at maximum effort are even significantly higher. The data come from different groups of athletes with different anthropometric data. Different estimation techniques (peaks instead of 80% robust means) were used in this example.

Sport-specific Actions

All sport-specific actions are summarized by the tested sports discipline/population in Table 2 and a comprehensive plot in our appendix. Overall, upper-body initiated activities yield lower compression values than lower-body, but the time expenditure seems to be higher.

Mixed Effect Models

The random intercept and slope regression models for standing, lifting, running and drop-jumps yield comparable results to the simple regression models shown in this publication - see our appendix.

Discussion

This study provides the mechanical spinal load in several activities of daily life and various sport-specific movements based on a mathematical model. This data can be used to compare activities, investigate relationships based on exercise intensity (e.g., velocity of running) and make predictions on new observations. Furthermore, the dataset can be enhanced with new incoming data using the same methodology.

Model Verification

Direct empirical validation of the model was not feasible, but there seems to be a reasonable agreement with in-vivo measurements. Intradiscal pressures were measured by Nachemson [19], Sato et al. [20], Wilke et al. and Takahashi et al. (2006) [21] in a healthy population. Rohlmann et al. [15,22,23]presented contact force data from measurements with instrumented implants. For standing, approximately 0.5 MPa (0.35-0.54 MPa) disc pressure is reported at L4/L5. For example, Wilke et al. (1999) [11] report a pressure of 0.48 MPa from direct intradiscal measurements in the intervertebral disc of the L4/L5 motion segment in the upright position. Considering a disc area of 12 cm² measured by MRI, this yields a compression force at L4/L5 of 575 N. Lifting a load of 20 kg showed an intradiscal pressure of 2.3 MPa and thus a compression load of 2,700 N. Applying the mathematical model to the same activities and a person with identical anthropometric data as the person examined by Wilke[11], a compression force of 550 N is calculated in the upright standing position and 2,600 N when lifting. Leskinen et al.[24] also calculated values between 3,000 and 4,000 N which is in line with our results. The compression forces measured with an instrumented implant when lifting a weight of 10 kg were shown to be 1,650 N[15].

In weightlifting (clean + jerk) with loads of >150 kg, maximum compression forces at L4/L5 of over 9,500 N were measured by Rohlmann et al.[15]. In this study, only 3 athletes performed cleans with jerks with 100kg and yielded an average estimate of about 8,500 N. Experience and lifting technique possibly play a crucial role, as the more experienced lifters had lower compression force than the linear model, based on less experienced lifters, predicts. Further, in powerlifting (285 kg), Granhed et al. (1987) [16] reported lumbar compression forces of over 15,000 N, although a very simplified static model was used for these calculations. The highest values from this study were about 12,500 N for long jumpers and 11,500 N for drop jumps with a height of 60 cm.

Thus, our model seems to be able to generate realistic data on spinal loading. Despite the limited assumptions and simplifications, the agreement found with the experimental data is quite good. The data generated by the model is likely to give a slightly conservative load estimate. With the given caution in the interpretation of the absolute values, we believe that the model yields reasonable predictions. Also, the data are based on investigations of different groups of athletes but were calculated with the same model throughout and are, thus, comparable with the given restraint.

Guideline

It is not surprising, that the estimates for several activities in sports overreach the threshold recommendation for lifting tasks among workers (3400N)[10]. While these activities are part of the daily \`working\` life of a professional athlete, overreaching is not avoidable. As a consequence, these recommendations do not hold for athletes and new recommendations should be developed. While these guidelines should incorporate the concept of tissue adaptation and a life-long development of physical and psychosocial resources to cope with these demands, physiological boundaries should be considered as well. For example, Brinckmann et al. [12] state, that the compressive strength is proportional to the product of bone density and the end-plate area of the vertebrae. On average, the female vertebrae are smaller and bone density decreases with age, but interindividual variability is very high among those factors[12]. Thus, sex and age may be taken into account for individual load estimation.

Practical Recommendation

The presented data can be used by practitioners to look up high demanding activities and use that information to balance out the high demands throughout training periods. While high peak loads for some activities cannot be avoided (e.g., lifters have to lift high weights to get better at lifting), the frequency and duration can be influenced. Thus, in the development of recommendations, such practical aspects should be considered. It is not feasible nor logical to resubstitute the frequency with a measure of energy expenditure like it was done in the NIOSH guidelines[10]. Lastly, also the duration of a single activity should be considered. We found that the duration of loading for many forms of stress associated with high loads in sports is relatively short, often as short as 50 to 100 ms (e.g. in running/sprinting). Nevertheless, low load repetitive movements or sustained muscle (co-) contractions can result in fatigue-related pain [25]

Low Back Pain and Injury

The role of mechanical load in the development of low back pain or injury is controversial. We argue that tissue adaptation plays an important role in adolescent athletes. It should be taken into account that the investigated activities are often performed by children and adolescents, whose musculoskeletal system often does not yet have the material properties and strengths of adults [26]. The compression forces differ considerably compared to adults, but adaptation processes of bone and connective tissue are generally slower compared to muscle tissue[27]. The rising performance level among adolescent athletes and the stress to compete with biologically accelerated but same-aged opponents probably lead to an increased risk of injury, considering the time of a young athlete's body to adapt to such high loads[28]. Again, to compensate for regular loads, monitoring based on training observations can be used to balance out spinal demanding activities throughout training periods.

For injury, there is a good body of resilient literature[29–31], that elaborates and provides sustained evidence, that any supporting and connective tissue will be damaged and destroyed, regardless of the biology, genetics, and psychosocial conditions present, or sex, age, degeneration, and activity level, when the mechanical load limit of only one tissue component is reached or exceeded. The injury may be spontaneous due to a current overload or gradual and accumulative after several repetitive microtraumas with submaximal loads. The failure criterion and limits may vary and depend on the loading history as well as the biochemical and biological environmental conditions of the tissue in question.

Limitations

Not all forms of stress and movement observed in sport and everyday life could be recorded for technical and organizational reasons. Nevertheless, the forms of stress taken into account appear to be representative and meaningful. Thus, a sustainable quantitative basis for a well-founded discussion of measures for the prevention of back pain and spinal injuries has been presented. In particular, indications of the necessity and advisability of developing and maintaining the musculature that supports and relieves the spine can be derived directly from the data presented. With the extreme stresses of many forms of athletic exertion appears. Further, estimates from this model cannot be directly compared with stress variables determined using other methods and models. Another limitation is, that gender-specific differences were not explicitly investigated partly due to missing data for participants gender. Lastly, as mentioned before, we provide no direct measurement of mechanical loading. The provided estimates are model-based.

Prospects

We propose that the results of our work and the data provided can be used to make model-based comparisons between physical activities and to make predictions on new observations. For example, the regression equation for constant running can be used to grossly estimate the normalized compression force with the formula *Normalized Compression* = 1.95 + 1.31 * velocity [m/s]. In future studies, we aim to further: 1) compare overhead activities in different sport disciplines and 2) examine longitudinal movement scenarios with a Bayesian prediction model. This model could be used for sports and daily activities but also in industrial settings, for risk predictions in adjunctions to existing tools[32].

Conclusion

We present a systematic examination of the mechanical spinal load in several activities of sports and daily life. This investigation can be inform the development of guideline recommendations for athletes, as the guidelines for workers cannot be applied. In conjunction, it is noteworthy that the kinetics of the spine in sport-specific activities are still rarely investigated. These findings are crucial in the development of youth athletes since to compensate for these high loads in the long run systematic preparation is necessary. Coaches can use the atlas of spinal mechanical loading to monitor spinal demands based on training observations and then progressively increase loading parameters. Also, given the verification of our approach with in-vivo data from the literature, this work can be considered as a quantitative basis for informed discussion of mechanical strain and prevention of back pain and injury.

Contributions

Conceptualization P.P.

Methodology G.-P.B, P.P., K.T., D.F., JF, KH

Biomechanical Modeling G.-P.B

Statistical Analysis R.S.

Investigation K.T., D.F., JF, KH

Writing – Original Draft R.S., K.T., D.F.

Writing – Review & Editing Everyone

Visualization R.S.

Funding

This study was conducted within the MiSpEx research network and funded by the German Federal Institute for Sport Science (BISp) [ZMVI1-080102A/11-18].

Declaration of interest

The authors declare no conflict of interest.

Supplemental materials

Data, code and the full report of our analyses are available in our repository: https://osf.io/vecgy/ DOI: 10.17605/OSF.IO/VECGY

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