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Abstract

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Methods Since the direct measurement is not feasible in sports activities, a mathematical model was applied to quantify spinal loading of more than 600 physical tasks in more than 200 athletes from several sports disciplines. The outcome is compression and torque (normalized to body weight/mass) at L4/L5.

Results The data demonstrate high compressive forces on the lumbar spine in 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.

Conclusion 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.

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ORIGINAL ARTICLE

The mechanical loading of the spine in physical activities

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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]. Injury may occur by exceeding the physical capacities. Cumulative loading can lead to microtraumata which weakens physical structures and, therefore, contribute to tissue failure (eg, ruptures, fractures) [7]. Hence, the loading profile (i.a., frequency, length, and intensity of exposure) is of high importance, which is also supported by findings of high prevalence of back pain among athletes and differences between sports disciplines [8]. A load-cycle interaction effect is suggested, which means that injury risk dramatically increases with high values in load and high repetition [7]. Moreover, this interaction effect is essentially driven by load [9]. Consequently, in-depth knowledge of the loading during different physical activities is crucial for effective risk assessment and prevention. However, to date, we lack specific guidelines for spinal loading profiles in athletes, nor do we have systematic investigations that might inform such.

Up to date, compressive forces of the lumbar spine were extensively researched via in-vivo measurements and mathematical modelling. Oxland [10, 11] and Dreischarf et al. [12] summarized the classical work of in-vivo measurements from Nachemson [13], Andersson et al. [14], Wilke et al. [15] and Sato et al. [16] for various daily activities. For example, intradiscal measurements at L4/L5 yield values of about 0.5 MPa for standing and 2.3 MPa for lifting [15]. High compressive forces when lifting heavy weights were shown by Granhed et al. [17] and Rohlmann et al. [18]. Lifting loads of 150 kg yielded compressive forces higher

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than 9500 N [18]. Powerlifters were shown to reach lumbar compressive forces higher than 15,000 N lifting 285 kg [17]. The estimates from in-vivo measurement procedures were commonly used to indirectly validate the modelling procedure [12]. Moreover, mathematical modelling can give insights into highly dynamic movements in various sports, where invasive measurements are hardly ethically justifiable. It has to be acknowledged that there are shortcomings in the modelling procedure, which are discussed in the review by Dreischarf et al. [12]. Also, it was shown that compressive forces alone are not sufficient to explain back injury as they could not reliably reproduce injuries like disc herniations in cadaveric studies [10]. However, a strategic evaluation of loads in sports is yet to be done to inform the pathophysiologic process of back injuries and can provide an important background for risk evaluation and prevention.

This study aims to estimate the mechanical loading of the spine via mathematical modelling for common physical activities and movements among many sports disciplines. This database will be useful in the development of tools or methods to prevent injuries. As an adjunct aim, we want to verify the estimated values by comparing them with the few available values from the literature.

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 as possible to competition level 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.

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 (Fig. 1). Ground reaction forces were recorded using force platforms embedded in the ground. Force data were recorded synchronously with motion data and sampled at 1000 Hz.

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 2000 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 [19]. This model assumes

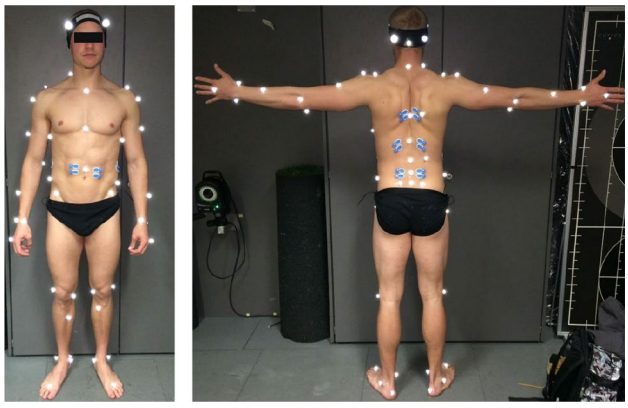

Fig. 1 Experimental setup of the 54 markers

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.

The contact force was calculated as the sum of the mechanical loads on the joint surface, which can in turn be decomposed into a compression force and a shear force component. In this model, we have omitted the facet joints due to the relatively small contact area. It is important to note, that this assumption may be more or less influential in different postures.

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 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.

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 stress the muscles involved (i.e., the square of the force related to the physiological cross-section) was chosen as the minimization criterion [20, 21]. Furthermore, the boundary conditions for the optimization were set as follows:

$0 \leq f_i^m \leq a_i^m$ and $0 \leq f_i^c \leq a_i^c$

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 joint, 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 mass* [kg])

- Compression [N] (f^c)
- Compression relative to body weight [AU] (f^c/*body weight* [N])
- Loading time [s or ms]
- "Spinal load" [AU]: Relative compression integrated over time as cumulative load drung the task

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/s^2). The parameters were determined in an interval at > 80% of the maximum of the compression force during the respective movement or posture.

Statistics

The mean and standard deviation are presented for each activity categorized by the tested population. Further, bootstrapped non-parametric 95% confidence intervals are calculated within each activity and each parameter. For certain activities (running, lifting, jumping, standing), additional simple linear regression, multilevel and spline models were fitted and compared. Some overhead activities will be extensively studied in a dissertation work and were thus excluded from the current analysis. Due to the explorative nature of this analysis, no formal significance test was applied. All computations were done in R v4.0.4 [22]. The full package list and the details of our analyses can be viewed in our reproducible R-Markdown script in our repository: https://osf.io/rnfam.

Results

Our dataset contains 637 observations from 248 participants. After filtering activities with only one observation and overhead activities, the final data set contained 578 observations from 236 participants, 18 groups (e.g., basketball, volleyball, hockey) and 67 investigated activities. Participant characteristics are shown in Table 1. The results of the studied activities are summarized in Table 2.

All sport-specific actions are summarized by the tested sports discipline in Table 2 and Fig. 2 illustrates the relationship between movement speed or initial energy and compressive forces at L4/L5 among different activities of daily living and sports. Further, an interactive web application provides the raw and summary data, and a comprehensive plot: https://robshavr.shinyapps.io/spinal-loading/. Overall, upper-body initiated activities yield lower compression values than lower-body, but the exposure time seems to be higher.The highest values in sports games were observed for intense lower pass actions in volleyball (dig: 5.2 times bodyweight ($CI_{95\%boot}$ [3.4; 6.9]), dive: 5.8 times bodyweight ($CI_{95\%boot}$ [4.9; 6.6]), change of direction (4.8 times bodyweight ($CI_{95\%boot}$ [3.4; 6.3]) and argentinian backhand (5 times bodyweight ($CI_{95\%boot}$ [4.1; 5.8]) in hockey and dribbling (5.5 times bodyweight ($CI_{95\%boot}$ [5.1; 5.9]) in basketball.

Table 1 Participant characteristics per cohort/discipline

Cohort/discipline	n	Body mass [kg] Mean (SD)	Height [cm] Mean (SD)	Female/male
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 from chair	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
Running	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
Jumping	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
Walking	10	71.1 (9.4)	174.1 (6.1)	4/6
Total	236	74.5 (9.4)	178 (8.9)	79/157

Mean: arithmetical mean, SD: standard deviation

Standing

When standing upright, the average estimated compressive 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 Fig. 3A). 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.

Lifting

When lifting a barbell of 10 kg, the average estimated compressive 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

Table 2 Outcomes for all activities tested grouped by cohorts tested

Cohort/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 0°	15	596 (184)	[508; 685]	0.9 (0.3)	[0.7; 1]	749 (348)	[575; 911]
Standing 30°	15	1271 (351)	[1094; 1436]	1.8 (0.6)	[1.5; 2.1]	1249 (348)	[1061; 1411]
Standing 90°	15	2195 (560)	[1921; 2461]	3.1 (0.9)	[2.7; 3.5]	1549 (348)	[1371; 1705]
Walking							
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 from chair							
Sit to stand	9	2384 (622)	[1986; 2780]	3.1 (0.9)	[2.6; 3.7]	712 (189)	[601; 826]
Pick sth. 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 one-hand	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]
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]
Change of direction	7	3166 (1246)	[2303; 4037]	4.8 (2.1)	[3.4; 6.3]	171 (89)	[116; 235]
Argentinian 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							
Throw	7	316 (129)	[224; 407]	0.4 (0.2)	[0.3; 0.6]	142 (56)	[108; 181]
Preparation	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]

Table 2 (continued)

Cohort/activity	n	Compression [N]		Normalized Compression [AU]		Time [ms]	
		Mean (SD)	$CI_{95\%boot}$	Mean (SD)	$CI_{95\%boot}$	Mean (SD)	$CI_{95\%boot}$
Long jump	5	12,456 (1359)	[11468; 13608]	15.7 (1.5)	[14.5; 16.8]	69 (10)	[62; 78]
Lifter, amateur							
Lateral lift 2 × 20 kg dumbell	11	1826 (346)	[1641; 2045]	2.5 (0.4)	[2.3; 2.7]	591 (119)	[523; 657]
Lift 10 kg barbell	20	2632 (479)	[2432; 2841]	3.9 (0.7)	[3.6; 4.2]	324 (131)	[271; 380]
Lifter, elite							
Clean 10 kg	11	2211 (358)	[2003; 2384]	3 (0.4)	[2.7; 3.2]	591 (119)	[527; 656]
Clean 20 kg	11	2867 (367)	[2644; 3047]	3.8 (0.3)	[3.6; 4]	591 (119)	[523; 656]
Clean 50 kg	11	4782 (772)	[4405; 5236]	6.4 (0.8)	[6; 6.9]	600 (121)	[530; 669]
Clean 80 kg	6	7979 (443)	[7652; 8281]	9.2 (0.5)	[8.8; 9.5]	540 (122)	[449; 631]
Clean 100 kg	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]
Running							
Running 2.5 m/s	20	3184 (670)	[2893; 3474]	4.7 (0.6)	[4.4; 5]	121 (26)	[111; 132]
Running 3.5 m/s	20	4775 (1006)	[4354; 5225]	7 (1)	[6.6; 7.4]	109 (23)	[100; 119]
Running 4.5 m/s	20	5681 (1167)	[5180; 6196]	8.4 (1.1)	[7.9; 8.9]	104 (22)	[95; 114]
Running 5.5 m/s	20	6079 (1248)	[5544; 6668]	8.9 (1.2)	[8.4; 9.5]	93 (20)	[85; 102]
Running 6.5 m/s	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 preparation	8	3574 (1523)	[2593; 4535]	4.7 (1.9)	[3.3; 5.9]	278 (120)	[220; 365]
Jumping							
Counter movement jump	20	3344 (507)	[3113; 3552]	4.7 (0.5)	[4.5; 4.9]	32 (15)	[26; 39]
Dropjump 20 cm	20	8359 (1267)	[7792; 8915]	11.7 (1.2)	[11.3; 12.2]	16 (8)	[13; 20]
Drop jump 40 cm	20	9613 (1457)	[8996; 10199]	13.5 (1.3)	[12.9; 14]	18 (8)	[14; 22]
Drop jump 60 cm	20	11,953 (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]
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

cohort of more experienced lifters, the estimated average force in lifting 10 kg is considerably lower (normalized compression: 3.0 $CI_{95\%}$ [2.7; 3.2]). Also, lifting 2 × 20 kg dumbbells laterally resulted in even lower normalized compressive force (2.5 $CI_{95\%}$ [2.2; 2,7]). The highest compressive forces in this category were observed when pushing a bobsleigh from professional athletes: 8976 N ($CI_{95\%}$ [7596.8; 10,264.4]), 10.8 ($CI_{95\%}$ [9.8; 11.7]) times body weight. In a linear regression model, the increase of compression per kg lifted in cleans is 75 N $CI_{95\%}$ [68; 82] and the predicted compression with no weight was 1352 N $CI_{95\%}$ [1018; 1695] (Fig. 3B).

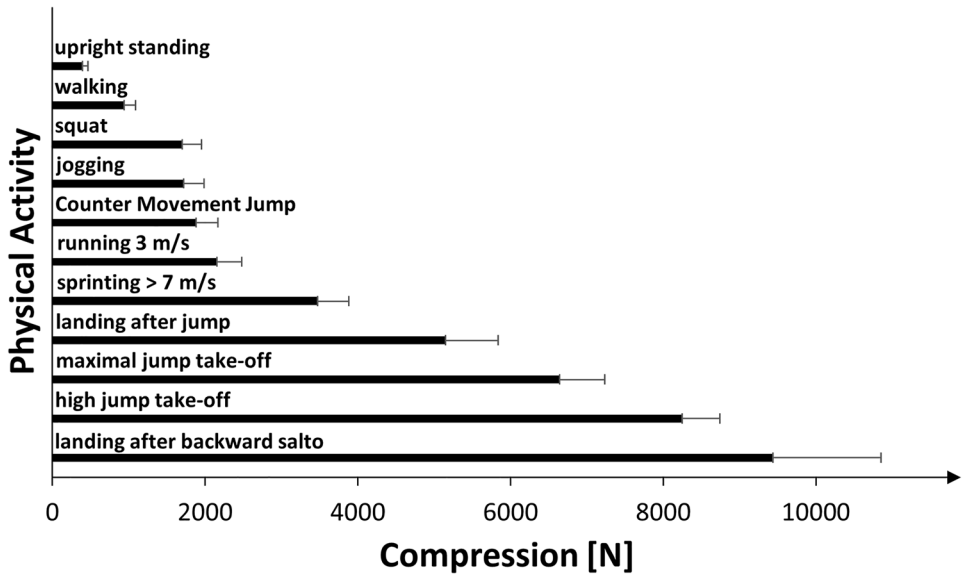

Fig. 2 Maximum compressive forces at L4/L5 during different physical activities. All activities were studied under training conditions in the laboratory. It is to be expected that the loads in the competition situation and at maximum effort are even significantly higher. The data come from different groups of athletes with different anthropometric data. Different estimation techniques (peaks instead of 80% robust means) were used in comparison to the rest of this article

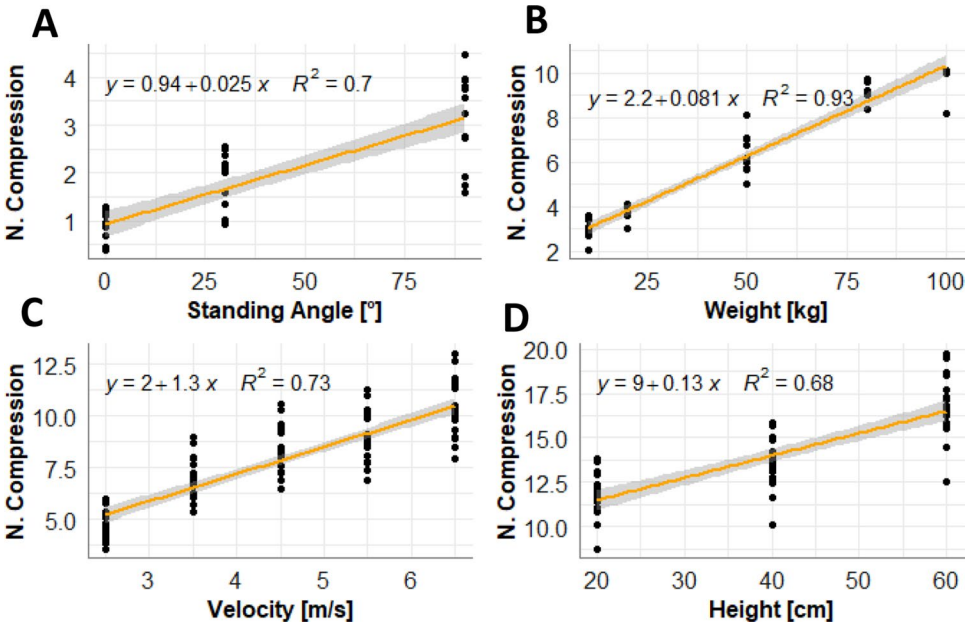

Fig. 3 Linear regression models for standing, lifting, running and jumping. Outcome (y-axis) is the normalized compression (compression [N]/bodyweight [N]). **A** Standing with trunk inclination, model: compression ~ trunk angle [°], **B** cleans with barbell, model: compression ~ lifted weight [kg], **C** constant running with different speeds, model: compression ~ running speed [m/s], **D** dropjumps from different heights, model: compression ~ drop height [cm]

Walking/running

When walking, the average estimated compressive 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 compressive 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 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 compressive force of 3113 N ($CI_{95\%}$ [2648; 3578]) or 4.6 ($CI_{95\%}$ [4.1; 5])—see Fig. 3C. 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 nonlinearity when increasing speed from 2.5 to 3.5 m/s. Moreover, the time spent at these forces decreases over time (see Fig. 4), 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 (< 100 ms) but—as in the nature of running—highly repetitive.

Jumping

When performing a countermovement jump, the average estimated compressive force at L4/L5 is 3343 N ($CI_{95\%}$

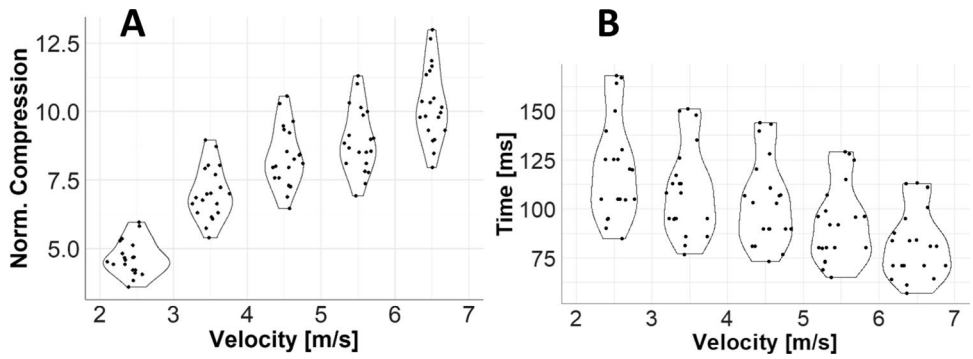

Fig. 4 Violin plots for running outcomes: normalized compression and time. **A** Increasing trend for compression with running speed. **B** Decreasing trend for time with running speed

[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 compressive 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 20 cm, 40 cm and 60 cm were tested. A linear regression model predicts for 20 cm drop-jump height an absolute compressive force of 8178 N ($CI_{95\%}$ [7550; 8806]) and relative to bodyweight 11.5 ($CI_{95\%}$ [10.9; 12.1])—see Fig. 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.

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 article. Details are provided in our analysis script within our online repository: https://osf.io/rnfam.

Discussion

This study provides valuable information on the mechanical spinal loading in several activities of daily life and sport-specific movements based on a mathematical model. The 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 and comparison with the literature

A direct empirical validation of the model was not feasible, but there seems to be a reasonable agreement with in vivo measurements from the literature [13, 16, 23].

In standing, Rohlmann et al. [18, 24, 25] and Wilke et al. [15] measured intradiscal pressure at L4/L5 of 0.5 MPa (0.35–0.54 MPa) and 0.48 MPa, respectively. Considering a disc area of 12 cm^2, this yields a compressive force of ca. 575 N, which is in line with our results. Our model predictions for the same person (75 kg) yields a compressive force of 550 N.

In lifting (20 kg), Wilke et al. [15] measured an intradiscal pressure of 2.3 MPa and thus a compression load of 2700 N, where our model predicts a load of 2600N. This is in line with Leskinen et al. [26], who calculated values between 3000 and 4000 N. The compressive forces measured with an instrumented implant when lifting a weight of 10 kg were shown to be 1650 N [18].

In weightlifting (clean + jerk) with loads of > 150 kg, maximum compressive forces of over 9,500 N were measured by Rohlmann et al. [18]. In this study, only 3 athletes performed cleans with jerks with 100 kg and yielded an average estimate of about 8500 N. Experience and lifting technique possibly play a crucial role, as the more experienced lifters had lower compressive force than the linear model, based on less experienced lifters, predicts.

In powerlifting (285 kg), Granhed et al. [17] calculated lumbar compressive forces of over 15,000 N, though 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 model yields rather conservative load estimates. 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.

Low back pain and injury

There is a good body of resilient literature [27–29], 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. The estimates from this study can inform the load/intensity aspect, which is highly important. However, frequencies should be taken into account in further research. Monitoring movements in sports and daily life in adjunction with estimated loads from this study might yield insights into the injury process. For example, this model could be used in an industrial settings in adjunction to existing tools [30].

Tissue adaptations play also an important role in the compensation of mechanical load in the development of low back pain or injury, especially 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 [31]. The compressive forces differ considerably compared to adults, but adaptation processes of bone and connective tissue are generally slower compared to muscle tissue [32]. 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 [33]. Again, to compensate for regular loads, monitoring based on training observations can be used to balance out spinal demanding activities throughout training periods.

Athletes experience high spinal loading on a daily basis. Thus, it is not surprising, that the estimates for several activities in sports overreach the threshold recommendation for lifting tasks among workers (3400 N) [34]. No recommendations for athletes are available yet and should be developed. 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. [35] 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 [35]. Thus, sex and age may be taken into account for individual load estimation.

Limitations

A limited number of movements 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. Further, estimates from this model cannot be directly compared with stress variables determined using other methods and models. The model assumptions can be also viewed as limitation, e.g. that ligaments do not produce force. Lastly, no direct measurement of mechanical loading is provided in this study.

Summary

We present a systematic examination of the mechanical spinal load in several activities of sports and daily life. This investigation can 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 for developing recommendations for (adolescent) athletes, since long-time adaptations are necessary to compensate high and frequent loading of the spine. With this database, spinal loading could be monitored by coaches or used in epidemiological research. 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.

Author contributions Conceptualization PP, methodology G-PB, PP, KT, DF, JF, KH, biomechanical modeling G-PB, statistical analysis RS, investigation KT, DF, JF, KH, writing—original draft RS, KT, DF, writing—review and editing everyone, visualization RS.

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Data availability Data, code and the full report of our analyses are available in our repository: https://osf.io/vecgy/.

Declarations

Conflict of interest The authors declare no conflict of interest.

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