Estimating Lumbar Spine Compression Using Markerless Motion Capture

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Abstract*— Compressive forces at the L5-S1 intervertebral joint are a contributing factor to low back pain, a leading cause of work-related musculoskeletal disorders. Estimating these forces in field settings and comparing them to safety limits can support ergonomic risk assessments. While musculoskeletal modeling platforms such as OpenSim provide accurate estimations of spinal loading, their reliance on expert operation and high computational demands limits their practicality in field settings. This study presents a biomechanical model that estimates L5-S1 compression force using joint angles derived from a markerless motion capture system, along with the user's body weight, sex, and the lifted load. A proof-of-concept evaluation was conducted with one participant performing symmetric lifting tasks at 5, 10, and 15 kg. The proposed method's outputs were compared to OpenSim estimates, showing close agreement in peak compression force for 10 and 15 kg conditions, with normalized peak estimation errors of 4 ± 2% and 2 ± 1% during lifting and 11 ± 5% and 5 ± 4% during lowering, respectively. Larger errors were observed for the 5 kg condition and during mid-movement for all load conditions, likely due to unmodeled muscle co-contraction and static assumptions. Overall, the method shows promise for accessible in-field ergonomic assessment of peak spinal loads during lifting and provides a foundation for evaluating interventions such as occupational exoskeletons in real-world settings.

Keywords—Markerless motion capture, L5-S1 compression force, Ergonomic risk assessment, Biomechanical modeling, Musculoskeletal simulation

I. Introduction

Low back pain is one of the most prevalent musculoskeletal disorders worldwide, and its impact on the working population is particularly severe [1], [2]. The burden of back pain extends beyond individual suffering, contributing significantly to reduced productivity, absenteeism, and early retirement [3]. Repetitive lifting and lowering, heavy material handling, awkward postures, and the development of performance-related fatigue are among the key risk factors for lower back pain [3], [4]. Various criteria are used to assess the limits and risks associated with repetitive lifting, including physiological, psychophysical, and biomechanical factors [5], [6]. Focusing on the biomechanical aspect, the L5-S1 intervertebral joint is a primary load-bearing segment of the spine and is particularly susceptible to injury due to the high compressive forces it experiences during lifting tasks [7]. Monitoring compressive force at the L5-S1 joint is therefore a key biomechanical measure for evaluating injury risk in occupational settings [6]. This metric can also be used in studies evaluating the effectiveness of interventions, such as modifications to task procedures or workplace environments, as well as the support provided by wearable assistive technologies, including exoskeletons [8].

Various methods have been developed to estimate this load [9], [10], [11], [12]. One of the most detailed approaches involves musculoskeletal modeling, which incorporates anatomical representations of body segments and muscles, subject-specific scaling, and dynamic simulations to estimate muscle activations and joint reaction forces [9]. The estimated intervertebral forces in these studies have shown strong correlations with in vivo implant data, underscoring the method's validity [13], [14]. However, this approach requires expertise in analyzing motion data and operating simulation tools such as OpenSim [9], and it involves high computational costs, which can limit its practical application in field settings.

To address the need for simpler yet informative tools, several analytical methods have been proposed [10], [11], [12]. The Hand-Calculation Back Compressive Force method estimates L5-S1 compressive force using inputs such as body weight, height, lifted load, horizontal hand distance from L5-S1, and torso flexion angle [10]. This model assumes a single equivalent back muscle with a sagittal moment arm of 6.9 cm for males and 6.6 cm for females [10]. Its outputs have been validated against the University of Michigan's 3D Static Strength Prediction Program™ , which estimates spinal loading at L5-S1 and L4-L5 levels using simplified anatomical representations of back and abdominal muscles [11]. Additionally, Potvin developed the Linked-Segment Biomechanical Model, which estimates L5-S1 compression force by summing the compressive effects of body segment weights and muscle forces [12]. This model requires information including user body weight, external hand load, trunk angle, and the horizontal moment arms of both the load and the upper body center of mass [12]. It assumes a fixed sagittal moment arm of 6 cm for a single equivalent back muscle [12].

Although many existing models estimate L5-S1 compression force using simplified biomechanical assumptions, they often require manual input of posture parameters and rely on task-specific measurements. With the growing availability of markerless motion capture technology [15], it is now possible to

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extract joint kinematics efficiently using a smartphone camera, without the need for specialized equipment or expert supervision. Therefore, developing a biomechanical modeling approach that estimates L5-S1 compression forces solely from joint kinematics obtained via a markerless system could simplify in-field task risk assessments for employers.

This study presented a biomechanical modeling equation to estimate L5-S1 compression force based on joint angles obtained from a recently developed markerless motion capture system, PoseChecker, combined with the user's body weight, sex, and the lifted load. The estimated values were evaluated against outcomes from OpenSim simulations, which served as the reference standard.

II. METHODS

A. Participant

For this proof-of-concept study, one female participant (age: 28 years, height: 168 cm, body mass: 63 kg) was recruited. Written informed consent was obtained after a full explanation of the experimental procedures. The study was approved by the University of Alberta Research Ethics Board (Pro00109264).

B. Task

The participant was instructed to lift a crate from ground level to pelvis height and then return it to the ground. The task was repeated using crate weights of 5, 10, and 15 kg. The selected weights reflect typical occupational loads and commonly used load intervals in biomechanics research [16]. No specific instructions were provided regarding lifting posture or speed. The order of the lifted weights was randomized, and three separate trials were performed for each condition.

C. Data collection

Motion data were captured using a nine-camera motion capture system (Vicon, Oxford Metrics, UK). The participant was fitted with 51 reflective markers as shown in Fig. 1, and motion was recorded at 100 Hz. Two markers were also placed on the crate to detect the instances of lifting and lowering. Ground reaction forces and moments were simultaneously recorded using two force plates (OR6-7, AMTI, USA) at 1000 Hz. All motion data were low-pass filtered using a bidirectional fourth-order Butterworth filter with a cutoff frequency of 6 Hz.

The markerless motion capture system tracks body movement and estimates joint angles by generating a skeletal model based on key points detected from captured video frames [15]. In this study, a smartphone camera (Galaxy S24, Samsung Electronics, South Korea) was positioned at approximately a 45° angle relative to the sagittal plane. This angle was selected to maximize visibility of joint motion while preserving sufficient depth information. An example frame from the lifting trial, showing the skeletal structure detected by the markerless system, is presented in Fig. 2.

D. L5-S1 Compression Force Estimation Using Musculoskeletal Modeling (OpenSim)

The musculoskeletal model used in this study was the new fully articulated thoracolumbar spine (FATLS) model, which comprises 78 segments, 93 degrees of freedom, and 552 musculotendon actuators [13]. Spinal force estimations produced by this model have been extensively validated against intradiscal pressure and instrumented implant data during both static and dynamic lifting tasks [13].

Motion was simulated in OpenSim for the scaled musculoskeletal model using the Inverse Kinematics tool. To estimate muscle activations and forces, the Static Optimization tool was employed to resolve the muscle redundancy problem by minimizing the sum of squared muscle activations, given the generalized net joint forces. External forces, including ground reaction forces and the lifted load, were provided as inputs for this tool. Similar to [17], the lifted load was modeled as external forces applied symmetrically to both hands, with half of the total weight assigned to each. The timing of load application was determined based on the position data of the markers placed on the crate. Muscle force-length-velocity properties were incorporated into the static optimization. Finally, the joint reaction Analysis tool was used to estimate joint reaction forces, including L5-S1 compression force. To ensure repeatability and reduce errors associated with manual input

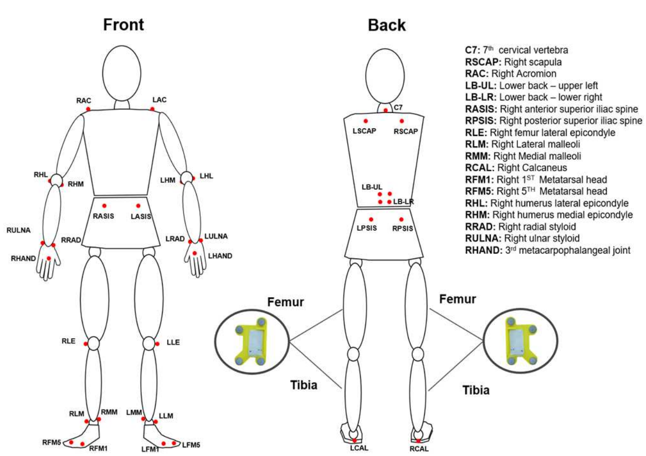

Figure 1. Placement of reflective markers on the body for motion tracking with the optical motion capture system.


Figure 2. Skeletal model captured by the markerless motion capture system during the lifting motion.

through the OpenSim's graphical user interface, the workflow was automated using custom MATLAB scripts (The MathWorks Inc., USA)

E. L5-S1 Compression Force Estimation Using Markerless Motion Capture (CF-ML)

The L5-S1 compression force was estimated based on the gravitational loads of the upper body segments and any lifted load, as well as the force generated by the erector spinae muscle. This simplified model assumes the erector spinae as the sole muscle generating the trunk extension moment and acting parallel to the spine. The compression force at the L5-S1 level was calculated using the following expression:

$$F = \left(\left(\frac{M_{L5-S1}}{ESMLA}\right) + \left((W_T + W_A + W_F + W_{Load}).\cos\theta_T + W_H.\cos(\theta_T + \theta_N)\right)\right).\cos 40^{\circ} \quad (1)$$

where W_T, W_A, W_F, W_H, and W_{Load} represent the weights of the trunk, arms, forearms, head, and the lifted load, respectively. θ_T and θ_N are the trunk and neck angles measured by the markerless motion capture system, as shown in Fig. 3. M_{L5-S1} denotes the net moment about the L5-S1 joint and is calculated based on the joint angles of the trunk, arms, forearms, and neck obtained from the markerless system. The moment computation also incorporates the weight, length, and center of mass location of upper body segments (Fig. 3), derived from anthropometric data provided by [18] based on the participant's sex and body mass. The term $ESMLA$ refers to the erector spinae muscle lever arm and is estimated using a regression model proposed by [19], which relates the lever arm to total body mass:

$$ESMLA\ (cm) = 3.469 + 0.025 \times M \quad (2)$$

where M is the body mass in kilograms. An orientation of 40° was assumed for the L5-S1 vertebral angle relative to the horizontal to project the net axial force along the spinal axis. This value was adopted based on previous studies reporting an average intervertebral angle of approximately 40° at the L5-S1 segment [20], [21]. It should be noted that the time intervals during which the crate was being carried were identified through visual inspection of the recorded video, and the contribution of the lifted load to the compression force was included accordingly through (1).

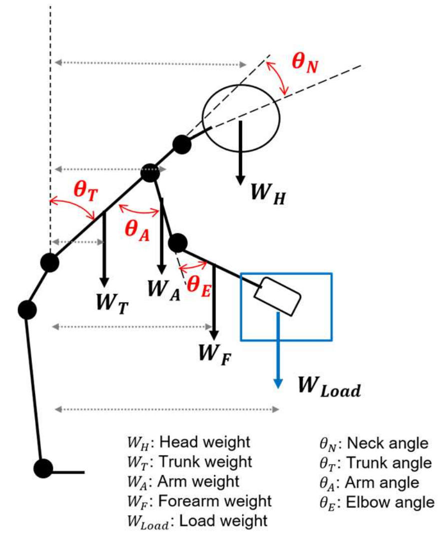

Figure 3. Illustration of the external loads and joint angles used in the L5-S1 compression force estimation using the markerless motion capture method (CF-ML).

F. Data Analysis

To enable direct comparison between CF-ML method and OpenSim results, all outputs were time-normalized to the duration of the lifting and lowering movement. For phase-specific analysis, the movement was divided into lifting and lowering phases, with the transition identified by the first local minimum in trunk angle following its initial peak, indicating the return to upright posture.

III. Results

Fig. 4 compares L5-S1 compression force estimates from the CF-ML method with reference values from OpenSim across three lifting conditions: 5, 10, and 15 kg. For the 10 and 15 kg conditions, CF-ML estimates generally aligned with the OpenSim reference in terms of peak magnitudes and the overall temporal pattern of compression force during lifting and lowering. However, noticeable discrepancies were observed during the transition to an upright posture and before initiating the lowering phase. In the 5 kg condition, larger deviations in peak magnitude were evident, particularly during the lifting phase.

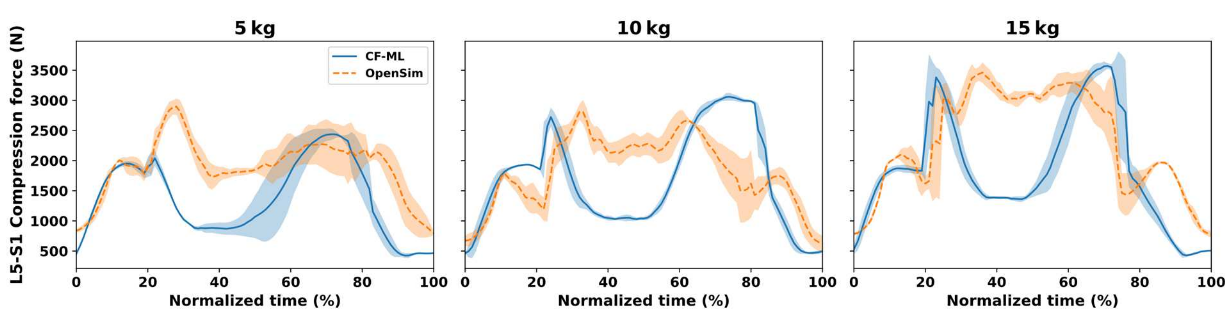

Figure 4. Comparison of L5-S1 compression force estimated using the markerless motion capture method (CF-ML) and the musculoskeletal modeling approach in OpenSim across three lifting conditions: 5 kg, 10 kg, and 15 kg. Shaded regions represent the standard deviation across three trials. Time is normalized to the duration of each lifting and lowering cycle.

Quantitative evaluation of these differences is summarized in Table I, with results reported separately for the lifting and lowering phases using normalized time thresholds of 34%, 41%, and 40% for the 5, 10, and 15 kg conditions, respectively. During the lifting phase, the normalized peak loading error for the 5 kg condition was $27 \pm 4\%$, while substantially lower errors were observed for the 10 kg ($4 \pm 2\%$) and 15 kg ($2 \pm 1\%$) conditions. The corresponding absolute peak error for the 5 kg condition reached 778.1 ± 135.3 N, indicating a relatively large deviation in peak magnitude compared to the reference values. In the lowering phase, the normalized peak loading error increased slightly for the 10 kg ($11 \pm 5\%$) and 15 kg ($5 \pm 4\%$) conditions relative to the lifting phase, whereas the error for the 5 kg condition decreased to $8 \pm 2\%$. Normalized root mean square error (RMSE) values remained relatively consistent across all load levels and movement phases, ranging from 28% to 34% of the OpenSim-estimated peak L5-S1 compression force.

IV. Discussion

This study presented a biomechanical modeling method for estimating L5-S1 compression forces based on joint angles obtained from a markerless motion capture system during dynamic lifting tasks. To evaluate the performance of this method, its output was compared to compression force estimates derived from OpenSim, which served as the reference.

Analysis of the estimated compression force time series revealed generally similar trends across methods prior to lift initiation (Fig. 4). However, noticeable discrepancies in force magnitude were observed during the phase in which the participant returned to an upright posture, across all lifted loads. Several factors may contribute to these differences. The CF-ML method relies solely on gravitational loads from upper body segments and the lifted load, along with a single equivalent extensor muscle force, all estimated based on upper body joint angles. This approach does not account for muscle co-activation, distribution of forces across multiple muscles, or passive intervertebral stiffness, all of which contribute to spinal loading and are modeled in OpenSim modeling. These simplifications likely contribute to the underestimation of compression forces during the post-lift ascent phase. Also, it should be noted that a previous validation study reported RMSE values of 6.5°, 12.9°, and 11.6° for the back, shoulder, and elbow joint angles, respectively, in the sagittal plane, when comparing this markerless motion capture system to a marker-based optical system [15]. Since the CF-ML highly relies on the joint angle measurements, errors in kinematic estimation directly affect the accuracy of the computed L5-S1 compression forces. We assessed the impact of ±6.5° back angle uncertainty on L5-S1 compression through an error propagation analysis. Peak load differences averaged 4.3%, 2.8%, and 3.1% for 5, 10, and 15 kg, with absolute differences of 106, 85, and 89 N, respectively. Furthermore, CF-ML method assumes static equilibrium at each time frame, while the static optimization approach incorporates the full dynamics of motion. It should be noted that the term "static" in this method used in OpenSim refers to the optimization formulation and does not imply a lack of motion-related dynamics. Previous studies have shown that incorporating inertial effects can substantially increase the estimated peak moments at the L5-S1 joint [22]. One study reported that inertial contributions can raise estimated compression forces by up to 60% compared to static calculations [23]. Additionally, it was found that under certain lifting conditions, peak moments may occur later in the movement cycle when dynamic effects are considered [22]. This observation aligns with the slight temporal lag in peak compression force seen in the OpenSim results compared to the CF-ML method estimates (see Fig. 4).

Peak compression force estimates from the CF-ML method closely matched OpenSim results for 10 and 15 kg loads during both lifting and lowering. However, as shown in Table I, a notably large normalized peak loading error was present for the 5 kg condition during the lifting phase. This discrepancy may be partly explained by the greater influence of muscle co-contraction on L5-S1 loading during lighter lifts, compared to the contribution from external loads and gross segmental weights, which are the sole inputs to the CF-ML method. This may reduce the accuracy of CF-ML in estimating peak compression forces under low-load conditions. The primary objective of the CF-ML method is to support ergonomic risk assessment in field settings by enabling comparison of peak spinal loads against established safety thresholds for L5-S1 compression. With the exception of the lifting phase for the 5 kg condition, all peak compression force estimates exhibited normalized errors of 11% or less. While these errors correspond to relatively small absolute differences, 308.6 ± 134.3 N or less,

TABLE I. Comparison of L5-S1 compression force estimation errors between the markerless method (CF-ML) and OpenSim reference values for three lifted loads (5 kg, 10 kg, and 15 kg), reported separately for lifting and lowering phases. Normalized values are expressed as a percentage of the corresponding OpenSim peak compression force.

	Lifting			**Lowering**		
	Peak loading error (N)	**Normalized peak loading error (%)**	**Normalized RMSE (%)**	**Peak loading error (N)**	**Normalized peak loading error (%)**	**Normalized RMSE (%)**
5 kg	778.1 ± 135.3	27 ± 4	28 ± 2	223.4 ± 61.3	8 ± 2	31 ± 6
10 kg	101.3 ± 48.5	4 ± 2	28 ± 3	308.6 ± 134.3	11 ± 5	34 ± 5
15 kg	85.5 ± 98.8	2 ± 1	28 ± 4	136.4 ± 94.4	5 ± 4	31 ± 3

caution should still be exercised when interpreting results that are close to established safety threshold values [6]. Also, further studies are needed to evaluate the accuracy of this method under lighter load conditions.

This study has some limitations. The CF-ML method simplifies muscle contribution to a single extensor acting parallel to the spine with a fixed lever arm, neglecting other muscles and variations in muscle moment arms during movement. It also assumes static equilibrium at each frame, which may underestimate compression forces, particularly during mid-motion. CF-ML was only evaluated for symmetric, two-handed lifting and requires manual identification of the lifting and releasing instances, which may introduce variability. Additionally, Higher errors in peak compression force under light loads likely reflect the greater influence of muscle co-activation relative to external loads. The study was conducted with a single participant, limiting generalizability. Future work should include more participants to validate the method, assess its accuracy for asymmetric tasks and light loads, and determine if a load threshold exists below which errors are unacceptably large. With a larger dataset, it may also be possible to develop load-specific correction functions to improve estimation accuracy, particularly under low-load conditions.

V. Conclusion

This study evaluated the feasibility of estimating L5-S1 compression forces using joint angles from a markerless motion capture system, combined with anthropometric, sex-specific, and load-related inputs. The method produced peak force estimates comparable to OpenSim for 10 and 15 kg lifts, with normalized errors under 11%. While larger errors were observed at 5 kg, the approach shows promise for in-field ergonomic risk assessments, where ease of use and accessibility are critical. Further research involving more participants and varied tasks is needed to enhance the method's accuracy and generalizability, particularly for lighter loads and asymmetrical movements.

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