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Assessing the accuracy of a wireless sensor system for estimating lumbar moments during manual lifting tasks considering the effects of load weight, asymmetry, and height

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ABSTRACT

This study assessed the accuracy of L5/S1 moment estimates calculated with an Inertial Motion Capture (IMC) system during an asymmetrical and variable height lifting task. The effects of load weight, asymmetry, and lifting height on estimates of lumbar moment have not been comprehensively considered in studies using IMC systems. Thirty-six participants engaged in tasks involving three loads, lifting heights, and trunk rotation angles. Lumbar moments were calculated using bottom-up and top-down biomechanical models. Gold-standard Optical Motion Capture (OMC) and Force Plates (FP) were used as the reference. A randomized block partially confounded design was used to compare the root mean square errors (RMSE) between the IMC and OMC-based reference estimates. The IMC system's estimated peak moments were 12%–13% lower than those estimated using the gold standard OMC-BU inverse dynamics, while the RMSE varied between 19 and 21 Nm. A Load*Height interaction was found; a trend was identified where the RMSE values increased as both the load and height levels increased. The angle did not show a significant effect on any of the tested scenarios. A close correspondence between the IMC and OMC-based moment estimates was established, with the load being the main factor affecting the differences between systems. The IMC system shows potential for use in occupational settings to capture data on the lumbar moments of workers, which could be utilized to assess ergonomic risk.

1. Introduction

Approximately 40% of all work-related musculoskeletal disorders (MSDs) in the US affect the back (U.S. Department of Labor, n.d.), and about 25% of workers report experiencing low back pain in the previous three months (Luckhaupt et al., 2019; Yang et al., 2016). Low back disorders have a significant economic burden worldwide (Bevan, 2015; Dagenais et al., 2008; Lo et al., 2021; Vos et al., 2012). Spinal loading resulting from biomechanical stress has been identified as a significant risk factor for low back pain (Coenen et al., 2013; da Costa and Vieira, 2009). Low back moment is one of the strongest predictors of low back disorder risk in manufacturing jobs (Marras et al., 2010).

Research conducted in a laboratory setting has investigated the effects of ergonomic interventions on the load placed on the spine, utilizing inverse dynamics analysis (Kingma et al., 1996) from laboratory-based instrumentation such as optical motion capture (OMC) systems, force plates (FP), and instrumented boxes or pressure sensing devices (Corbeil et al., 2019; Hwang et al., 2009; Plamondon et al., 2010, 2014). OMC is predominantly regarded as the benchmark or "gold standard" in recording the position of body segments in a 3D space, with recent reports in the literature of errors below 200 μm for 97% of a 135 m^3 capture volume (Aurand et al., 2017). The application of OMC and FP methodology is commonly used in laboratory-based ergonomic studies due to its accuracy on kinematics and kinetic estimates when using biomechanical and musculoskeletal modeling (Faber et al., 2020; Koopman et al., 2018; Larsen et al., 2020; Nail-Ulloa et al., 2021, 2024; Skals et al., 2021). Despite their accuracy, OMC and FP systems are impractical for work environments (Faber et al., 2011). To address this

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limitation, some wearable measurement systems have been developed for ambulatory assessment of backloading (Freitag et al., 2007; Marras et al., 2010), but they can be bulky and not practical for use in highly dynamic conditions. Inertial motion capture (IMC) systems consisting of small inertial measurement units (IMUs) have shown promise in assessing spinal loading, such as L5/S1 moments during trunk bending (Faber et al., 2016). Combining IMU sensors and force-sensing shoes has led to lumbar moments estimates within 10–20% of peak extension moments for manual lifting tasks (Faber et al., 2020). However, force-sensing shoes can be obtrusive in a real work environment, inviting us to consider an approach solely using IMUs. Additionally, the effects of the interactions between loads, lifting heights, and the asymmetry of a lift have not been comprehensively assessed with a model derived from IMUs. Previous studies have provided insights into the impact of individual factors such as load weight and lifting height on spinal loading (Hoozemans et al., 2008; Norasi et al., 2019). However, there is a lack of comprehensive understanding that integrates these factors with lifting asymmetry, particularly in examining how these variables influence the differences between observations from OMC and IMC systems. This gap is of significant relevance, especially when considering the practical application of IMC systems in dynamic work environments and their comparison with the gold standard OMC systems.

The current study has two main objectives. First, it aims to determine if L5/S1 moments can be accurately assessed using an IMC system during lifts that vary by three primary factors: the weight of the load, degree of asymmetry (angle), and height associated with the lifting and lowering tasks. These assessments were compared against bottom-up (BU) and top-down (TD) laboratory models constructed using a "gold-standard" OMC and FP methodology. Second, the study analyzes the effect of these controlled variables—load, angle, and height—on the error of the IMC-based system in estimating lumbar moments.

2. Methods

2.1. Participants

A total of 36 participants were recruited for the study, with an equal number of males and females, aged between 19 and 55 years old, and with a mean height of 173.54 cm (±7.5 SD) and a mean body mass of 72.78 kg (±12.1 kg SD). The participants were recruited from the Auburn University student body and the local Auburn/Opelika community. To be eligible, participants had to meet specific inclusion criteria, including 1) having no history of physician-diagnosed MSDs, injuries, or low back surgeries, 2) no low back pain in the past six months, and 3) no physician-diagnosed neurodegenerative disorders that may affect movement, such as Parkinson's disease or multiple sclerosis, among others. The study was approved by the Institutional Review Board of Auburn University (Protocol # 16–211 MR 1606). Details about the participants' anthropometry are provided in Table 1.

2.2. Participant preparation and data collection

Per established protocols, the research team collected participants' anthropometric measurements, including weight, height, length, and circumference of various major body segments (Haslegrave and Pheasant, 2018).

The participants were equipped with 17 IMU sensors (MVN Awinda, Xsens Technologies B.V., Enschede, the Netherlands); these IMU sensors are wireless, small, battery-powered, and capable of measuring and storing data on linear acceleration, angular velocity, and magnetic field. A combination of elastic neoprene straps and hypoallergenic athletic tape was used to secure the sensors. In addition, a full-body protocol of 50 reflective OMC markers was used. These markers were placed at specific reference points on the body to serve as the reference "gold standard" for body segment position. The reflective marker trajectories were recorded by 16 cameras (Optitrack Prime 13, Natural Point, Inc.). The primary marker locations were defined using the full-body Plug-In Gait 39-marker set (Vicon®, 2002), while 11 additional markers were used to keep track of some segments that may be obstructed during lifting tasks due to the nature of the movement. These 11 markers were placed on the inside of the elbows, ankles, knees, top of the feet, left and right sides of the pelvis, and between the sternum and clavicle markers (see Fig. 1). The information obtained from the IMU sensors and anthropometric data was utilized to build a rigid link biomechanical model. This model was then evaluated against the full-body Plug-In Gait OMC model, using the data gathered from the reflective markers. An example of the participant setup and OMC and IMC avatars is shown in Fig. 2. Two AMTI force plates (Advanced Mechanical Technology Inc., Watertown, MA) operating at a sampling frequency of 1000 Hz were used to collect ground reaction forces.

2.2.1. Systems calibration

The OMC system underwent calibration before each session of data collection, which involved defining the capture volume and establishing the origin and orientation of the global coordinate system. Before commencing the experimental procedure, participants were required to complete manufacturer-defined calibration protocols for both the IMC and OMC systems. Segment calibration was necessary for the IMC system to align the motion trackers to the participant body segments

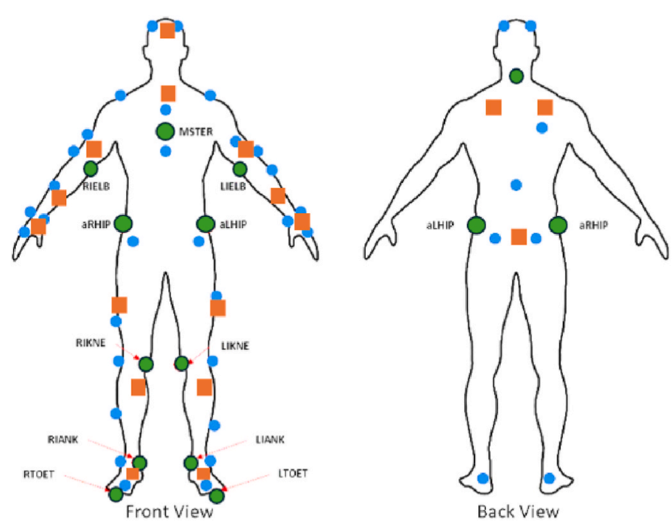

Fig. 1. Participant full-body setup, showing the modified Plug-in Gait marker configuration and IMU sensor placements. The diagram includes standard markers (blue circles), additional markers (green circles), and IMU sensors (orange squares).

Table 1
Summary of participants' anthropometry.

Mean	Body mass (kg)	Body height (cm)	Shoe length (cm)	Ankle height (cm)	Ankle width (cm)	Knee height (cm)	Knee width (cm)	Elbow width (cm)	Shoulder width (cm)	Shoulder height (cm)	Hip height (cm)	Hip width (cm)	Arm span (cm)
Male	79.6	178.4	26.0	7.2	9.5	48.2	10.2	7.2	35.8	146.2	93.0	24.8	178.4
Female	65.9	168.7	24.2	6.9	6.4	47.9	9.3	9.7	32.8	137.9	91.1	24.3	165.8

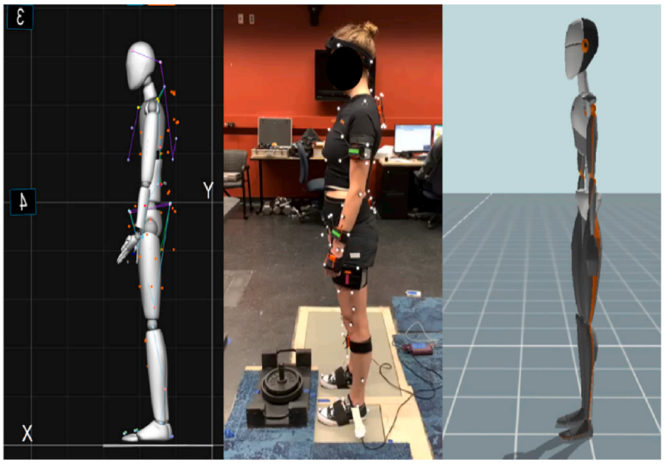

Fig. 2. Side-by-side comparison: (Left) OMC model avatar, (Middle) Participant, (Right) IMC model avatar.

(Xsens®, 2021). To calibrate the system, participants assumed a neutral posture (N-pose) for a brief period, then walked straight forward and returned to the starting point. For the OMC system (NaturalPoint®, 2018), segment calibration required the participant to adopt an "A" pose (as depicted in Fig. 1) to record a static trial. The static trial was subsequently associated with the dynamic trials for each participant, facilitating further biomechanical analysis.

2.2.2. *Data synchronization and processing*

The OMC software (Motive™) was defined as the primary system to synchronously capture data from the force plates, OMC, and IMC. The OMC data was captured at a sampling rate of 120 Hz, while the IMC data was sampled at 60 Hz. The OMC signals were down-sampled offline to 60Hz, and the reflective markers' trajectory gaps were filled using the cubic interpolation methods (Xu et al., 2010). A bi-directional low-pass filter with a second-order Butterworth filter at 6Hz was used for the OMC system (Karatsidis et al., 2016) and 5Hz for the IMC system (Faber et al., 2020) to filter the forces and kinematics.

2.2.3. *Biomechanical modeling*

Biomechanical models using inverse dynamics were employed to analyze the data. The bottom-up approach was exclusively applied to the OMC model, whereas the top-down approach was utilized for both the IMC and OMC model-analyses. In OMC systems, segment positions are determined based on the absolute positions of reflective markers placed on anatomical landmarks. In models like the Plug-in Gait segments are defined by capturing the 3D positions of the markers, and joints are calculated based on the spatial relationships between markers (Davis et al., 1991). The biomechanical model from the OMC data consisted of 15 segments (hands, forearms, upper arms, thighs, shanks, feet, pelvis, thorax, and head) and was developed in Visual3D software (C-Motion, Rockville, MD). The guidelines used to determine the location estimate for L5/S1 (Dumas et al., 2007; Reed et al., 1999) are based on proportions of the anterior superior iliac spine (ASIS) distance. This model was validated in a previous study (Nail-Ulloa et al., 2021).

The biomechanical model for the IMC system was also developed in Visual3D and consisted of the same 15 segments (C-Motion®, 2020). In IMC systems segment orientations are measured and then used to estimate the position of each segment by stacking them sequentially, assuming fixed segment lengths determined during calibration (Roetenberg et al., 2009).

The external loads in the top-down calculations were divided equally between the hands, with half the weight assigned to each hand segment (Marklin et al., 2024). To determine the moments of load transfer, the team recorded videos using digital cameras positioned from the sagittal plane to capture the instances when the loads made contact with the subjects. These videos were then synchronized offline with the corresponding trials to identify key moments of load transfer. Additionally, a 10-frame window moving average was applied on the trials to smooth the effect of load transferring.

2.2.4. *Experimental design*

The analysis involved calculating the total low back moments obtained by the vector summation of the time series of the L5/S1 moments (in Nm) from each plane (X, Y, Z) and estimating the peak moments and root mean square errors (RMSE). The 36 participants completed simulated manual material handling tasks in a laboratory environment, where kinematic and kinetic data were collected. The duration of the tasks varied from 10 to 30 s, depending on the participant's speed. Each participant performed three testing trials with a 2-min rest period between them.

Participants were instructed to lift boxes from the ground level to a shelf and lower them back to the ground. The lifting tasks varied randomly by weight, angle, and height: the boxes weighed 10 lbs. (4.54 kg), 20 lbs. (9.07 kg), or 30 lbs. (13.61 kg); they were lifted both symmetrically and asymmetrically (i.e., 0°, 30°, and 60° from the sagittal plane to the left); and they were lifted to one of three different shelf heights (60 cm, 100 cm, and 140 cm). The tested combinations were labeled using the following terminology: LXX (XX = load of 10, 20, or 30 lbs.), AYY (YY = angle of 0°, 30°, or 60° asymmetry), and HZZ (ZZ = height of 60, 100, or 140 cm). For example, L20_A30_H140 represents a 20-pound load picked up from an asymmetry angle of 30° and lifted to a height of 140 cm. Each participant completed three random combinations of load, angle, and height out of the 27 possible combinations described in Table 2. Loads and heights were randomized within levels of angle to ensure variability and reduce order effects. With 36 participants each performing three different testing trials, this resulted in a total of 108 different trials, divided into lifting and lowering phases.

2.2.5. *Data analysis*

To compare the systems' agreement levels, a combination of boxplots, Kernel Density Estimate (KDE) plots, simple linear regression analyses, and Bland-Altman plots was employed. KDE plots were utilized to visualize the distribution of errors between the IMC and OMC-BU systems, offering a smooth estimate that identifies patterns and variations crucial for assessing the IMC system's accuracy and consistency. Simple linear regression analyses were conducted to quantify the degree of association between the IMC and OMC-BU measurements, helping to identify any systematic deviations. Bland-Altman plots were used to evaluate the agreement between the IMC and OMC-BU measurements. A 3^3 randomized block partially confounded design (RBPF) with blocks of size three was used to compare the RMSE of the estimated moments. Confounded factorial designs are particularly appropriate if an interaction is expected to be negligible. The interactions can be confounded with groups, reducing block size without sacrificing power in evaluating the treatments (Kirk, 2014). Participants were divided into four groups

Table 2
Different combinations of load, height, and angle.

Load	Height		
	60 cm	100 cm	140 cm
10 lbs	0° 30° 60°	0° 30° 60°	0° 30° 60°
20 lbs	0° 30° 60°	0° 30° 60°	0° 30° 60°
30 lbs	0° 30° 60°	0° 30° 60°	0° 30° 60°

based on sex, with the first two groups consisting of male participants and the last two groups consisting of female participants. This grouping was designed to control for between-subject variability, particularly sex-related differences, which could influence the outcomes. The significance of the groups in our study indicates meaningful variation between them, justifying this approach. Although sex was evaluated, no statistically significant differences or large effect sizes were observed, and thus, these results were not reported.

The primary experimental factors (Load, Angle, Height) and their potential interactions were analyzed to identify significant differences. A type I error rate of 0.05 was used for all tests. Pairwise comparisons were performed using Tukey's test to analyze the RMSE with the main effects and interactions found to be significant. Effect sizes were calculated to measure the strength of the relationship between variables. Omega-squared (ω^2) is recommended for complex designs with multiple variables (Kirk, 1996). Segment length measurements were extracted for the right foot, shank, thigh, hand, forearm, upper arm, pelvis, and thorax for both biomechanical models. Next, OMC and IMC-based segments were analyzed using paired tests. All statistical analyses were performed using Python 3.7.

3. Results

This section is divided into two categories: first, comparing the moment estimates across all modeling and calculation methods; second, the results of the RBPF design and post hoc analyses, which delve into the impact of the variables of interest. Fig. 3 illustrates a subject performing a trial, Fig. 4 displays typical moment estimates associated with the lifting (A) and lowering trials (B). The figure includes the Bottom-Up Optical Motion Capture-based model (OMC-BU), the Top-Down Optical Motion Capture-based model (OMC-TD), and the Top-Down Inertial Motion Capture-based model (IMC-TD)—both figures from the lifting and lowering show two distinct peaks. The higher peak generally corresponds to picking up or setting down the load from and to the floor level. The other peak typically arises when placing or removing the load from the shelf. The results for OMC-TD can be found in Appendix 1.

3.1. Comparison of moment estimates

Table 3 displays the peak moment estimates, which show an average of a 13% difference between the peaks from the IMC-TD model and the OMC-BU (GS). Fig. 5 further supports these trends by displaying the KDE plots of the peak moment observations' distributions.

In Fig. 6, the Bland-Altman plots for the peak moments reveal a mean difference of 25.96 Nm for the lifting task (limits of agreement ± 38.47) and 22.22 Nm for the lowering task (limits of agreement ± 38.80) when comparing OMC-BU and IMC-TD.

Fig. 7 illustrates the results of the simple linear regression analysis for the tasks. For the comparison between OMC-BU and IMC-TD, the regression line equation suggests a slope of 0.96 (95% CI: 0.87–1.04) for the lifting task and 0.91 (95% CI: 0.82–1.01) for the lowering task. The intercepts for lifting were 33.47 Nm (95% CI: 33.38–33.55), and for lowering were 37.05 Nm (95% CI: 36.95–37.14).

Fig. 8 showcases the RMSE averages for the different lifting variables. The Load factor shows a trend of increasing the RMSE values as the load level increases.

3.2. Lifting variables and segment length

The analysis compared the RMSE for the lifting and lowering tasks, and an exploratory analysis on major segment length differences.

The ANOVA revealed significant differences between the participant groups across all scenarios, as presented in Table 4. A significant interaction between Load and Height was observed for the lifting ($p = 0.003$) and lowering tasks ($p = 0.002$) for the comparison of OMC-BU vs. IMC-TD. As illustrated in Fig. 9, the interaction suggests that the effect of load on RMSE varies depending on the height level. Specifically, while RMSE increases with load across all heights, the magnitude of this increase is not uniform, indicating a differential effect of load depending on the lifting height.

The lengths of the different segments were compared (Table 5), with the Thorax and Right Thigh having the higher differences, as seen in Fig. 10. Paired t-tests showed that the thorax ($p < 0.001$, Cohen's $d = 2.262$) and the thigh ($p < 0.001$, Cohen's $d = -1.515$) are significantly different.

4. Discussion

The current study assessed the efficacy of an IMC system in estimating L5/S1 moments across varying load, height, and asymmetry angle configurations. The performance of a state-of-the-art laboratory system—comprising an OMC and FPs and a bottom-up OMC model—was benchmarked against a top-down IMC model. For internal validation of the OMC system, L5/S1 moments were also determined using a top-down OMC model. The findings indicate that the IMC system effectively estimates peak moments, underscoring its potential for biomechanical field assessment.

5. Accuracy of moment estimates

Overall, the IMC-based model estimates were 12%–13% lower than the peak moments estimated using the OMC-BU inverse dynamics. These

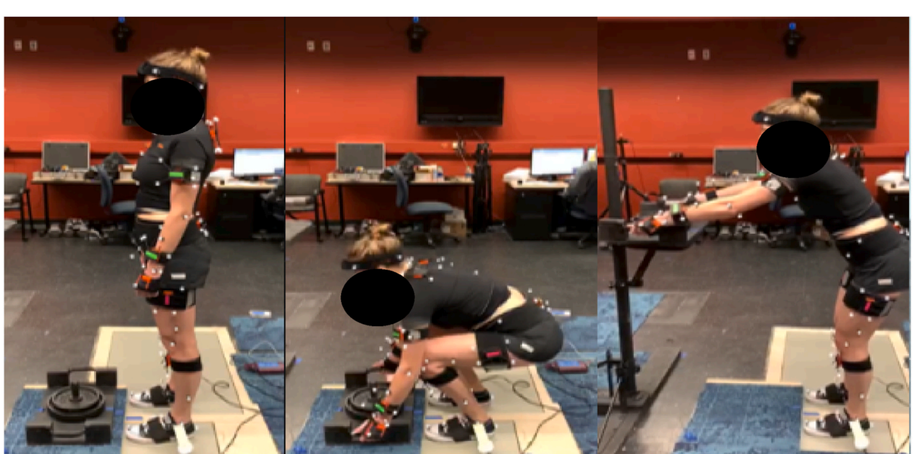

Fig. 3. Participant during a trial at rest (Left), initial lifting (Middle), and placing the load on the shelf (Right).

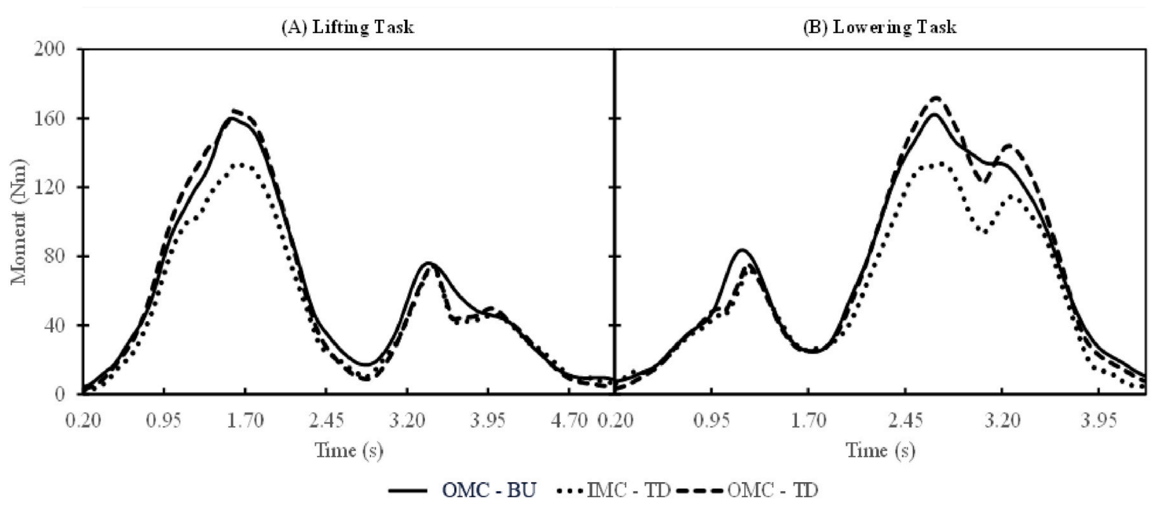

Fig. 4. Resultant L5/S1 moment from a subject for the combination L20_A00_H100 L5/S1 during one of the liftings (A) and lowering (B) trials.

Table 3
Summary of measures of the moment estimates for the different modeling approaches, with values in parenthesis, indicating the percentage difference from the Gold-Standard OMC-BU.

Summary measure	Peak Moment		RMSE	
	Peak OMC-BU	Peak IMC-TD	OMC-BU OMC-TD	OMC-BU IMC-TD
Lifting (Nm)				
Mean	197.59	171.63 *(-13%)*	16.03	19.07
SD	45.09	42.5	6.4	5.94
Max	320.25	295.35	39.64	34.74
Min	110.91	97.66	4.92	8.2
25th percentile	161.94	139.39	11.44	15.28
50th percentile	195.02	166.87	15.5	18.05
75th percentile	231.32	203.23	19.3	23.23
90th percentile	258.91	224.61	25.33	26.86
Lowering (Nm)				
Mean	188.47	166.24 *(-12%)*	16.95	21.38
SD	41.1	39.74	6.79	7.24
Max	284.01	291.89	43.3	41.72
Min	96.55	98.14	4.22	6.18
25th percentile	157.62	136.41	12.28	16.05
50th percentile	186.59	161.45	16.06	20.74
75th percentile	219.73	195.06	19.64	25.58
90th percentile	243.39	220.55	25.09	30.46

results align closely with findings from similar studies (Faber et al., 2016, 2020; Koopman et al., 2018). The linear regression analysis of peak moment estimates, as presented in Fig. 7, indicates that a total moment of 250 Nm obtained from the IMC system for lifting and lowering tasks corresponds to 273.47 Nm and 264.55 Nm, respectively. This relationship offers valuable insight into potential equivalencies when utilizing the IMC system without force platforms. The results suggest that distributing the load between the participants' hands is promising for similar lifting conditions. We recognize that tools have different shapes and weight distribution in industry. Nevertheless, this approach is worth exploring when analyses must be simplified.

A recurrent pattern emerged when comparing moment estimates across trials: the IMC system's estimates were consistently lower than those of OMC-based models, as detailed in Table 3. Previous research has noted similar patterns (Faber et al., 2016). This discrepancy can likely be traced back to the distinct methodologies the two systems employ in tracking and defining body segments. While OMC-based models rely on bony landmarks to estimate segment lengths — potentially yielding more accurate results — IMC-based model estimates draw on the participant's segment orientation and anthropometric data, such as height, arm span, shoulder, and hip heights. With this method, errors accumulate in segment positions along the kinematic chain, particularly at the distal segments, due to the propagation of orientation errors (Zhou and Hu, 2008). These cumulative errors can result in inaccuracies in the estimated position of the load, potentially contributing to the underestimation observed in the IMC system.

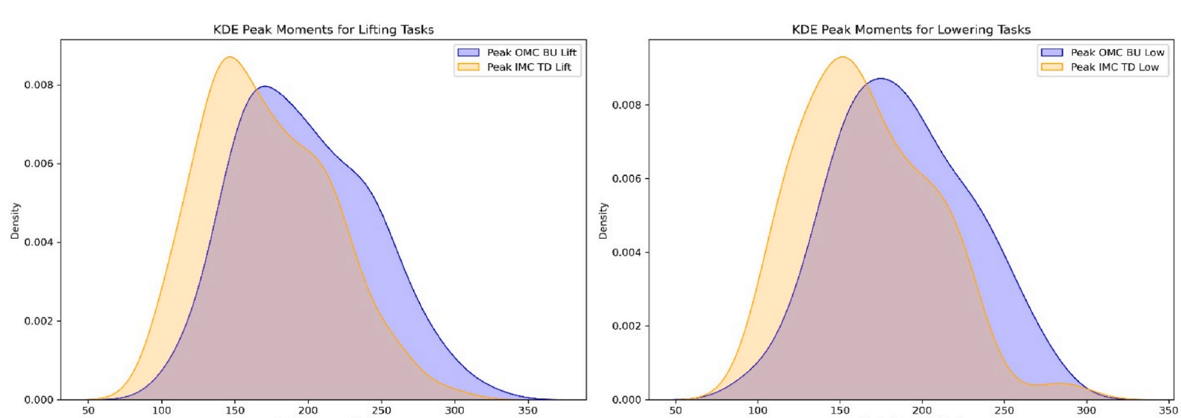

Fig. 5. KDE plots for L5/S1 Peak Moments for the Lifting (Left) and Lowering (right) tasks.

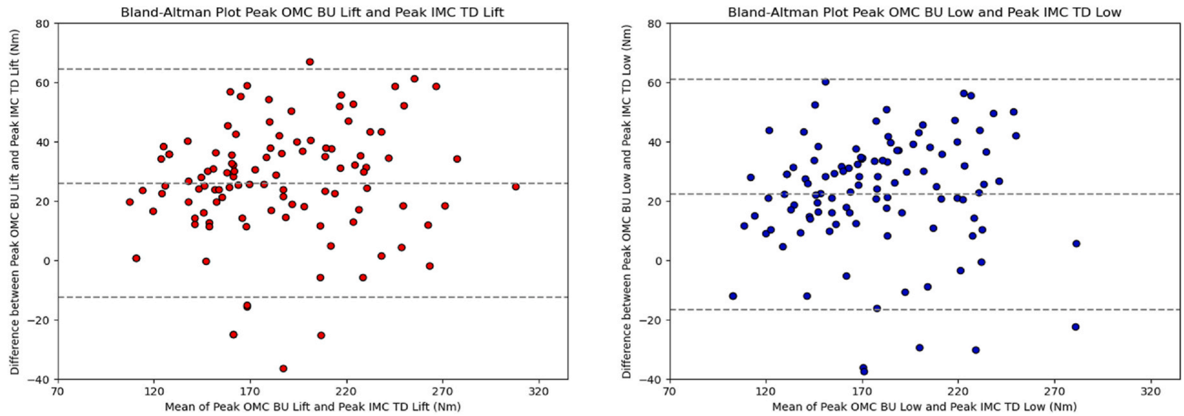

Fig. 6. Bland-Altman plots for Peak Moments for the lifting (left) and lowering (right) tasks, with 108 data points representing 36 participants performing three different combinations of load, angle, and height.

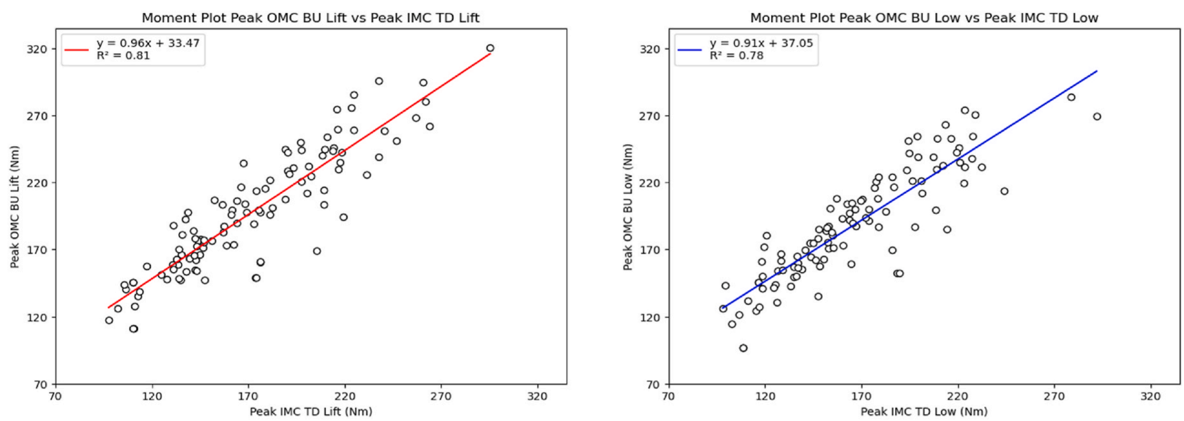

Fig. 7. Scatter plots for Peak Moments for the lifting (left) and lowering (right) tasks, with 108 data points representing 36 participants performing three different combinations of load, angle, and height.

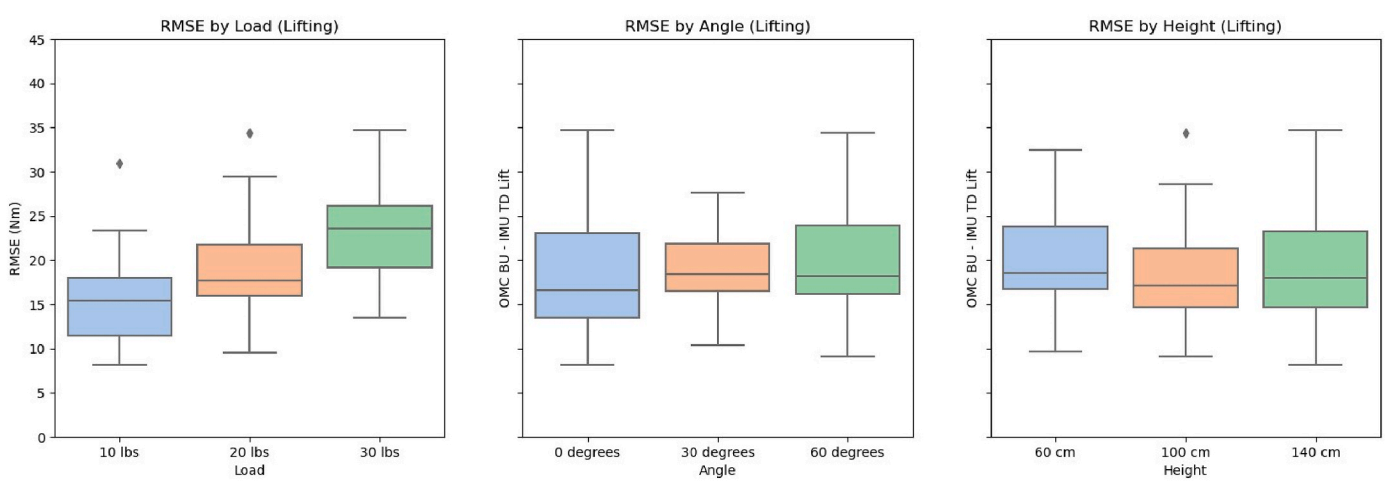

Fig. 8. L5/S1 RMSE boxplots for the Load (left), Angle (middle) and Height(right) for lifting trials.

During our analysis, we noted that specific segment lengths, especially that of the thorax, were consistently underestimated by the IMC system, often by an average of 6 cm. Intriguingly, the underestimation in the IMC-based model starts from the pelvis and extends upwards; however, the IMC model overestimated thighs, shanks, and foot segments. This discrepancy in segment lengths (or lever arm) may play a significant role in the observed lower moment estimates from the IMC model compared to the OMC model's TD and BU approaches. Further research is needed to investigate these differences in segment length estimation and their direct implications on moment estimates.

Another possible factor that could influence the variability among the models in this study is how the thorax is defined for TD calculations. As the trunk is not a rigid body, using a single rigid segment to model it may introduce potential sources of error. Nonetheless, it is feasible to

Table 4
ANOVA results testing the effects of Loads (A), Angle (B), Height (C), and interactions for the RMSE of the moments at L5/S1 during the lifting tasks. Significant effects ($p < 0.05$) are indicated in ***bold/italics.*** Large effect sizes ($\omega^2 > 0.14$) are shown in **bold**.

Factors (df, df Error)	RMSE OMC BU - IMC TD	
	p-value	Effect Size ω^2
Lifting		
Groups (3, 46)	***0.088***	0.103
A - Load (2, 46)	***<0.001***	**0.671**
B - Angle (2, 46)	0.448	−0.003
C - Height (2, 46)	0.072	0.027
A*B (4, 46)	0.613	−0.010
A*C (4, 46)	***0.003***	0.114
B*C (4, 46)	0.575	−0.008
A*B*C (8, 46)	0.390	0.005
Lowering		
Groups (3, 46)	***0.012***	**0.213**
A - Load (2, 46)	***<0.001***	**0.547**
B - Angle (2, 46)	0.412	−0.002
C - Height (2, 46)	0.416	−0.002
A*B (4, 46)	0.560	−0.011
A*C (4, 46)	***0.002***	**0.181**
B*C (4, 46)	0.190	0.028
A*B*C (8, 46)	0.914	−0.056

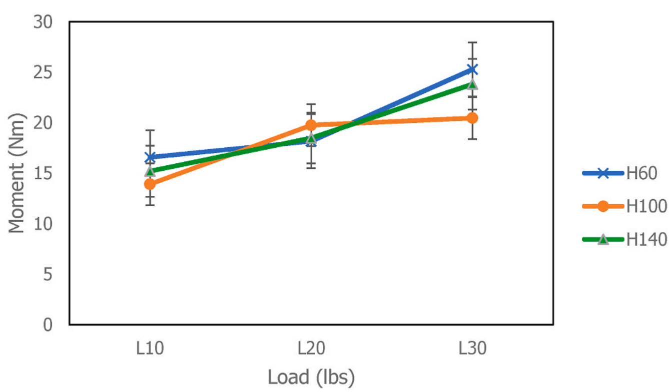

Fig. 9. Interaction plot for Load*Height on RMSE (OMC-BU vs. IMC-TD) during lowering tasks, error bars denote the standard error.

Table 5
Segment length comparison between OMC and IMC biomechanical models.

Segment	Average Length OMC (m)	Average Length IMC (m)	Average Difference (m)	SD of Difference (m)
Right Hand	0.07	0.07	0.01	0.01
Right Forearm	0.24	0.23	0.01	0.01
Right Upper Arm	0.29	0.29	0.01	0.02
Thorax	0.46	0.39	0.06	0.03
Pelvis	0.10	0.09	0.01	0.01
Right Thigh	0.41	0.45	−0.04	0.03
Right Shank	0.39	0.42	−0.03	0.04
Right Foot	0.14	0.15	−0.01	0.01

employ a defined rigid trunk in loading investigations at the lower levels of the spine (Ignasiak et al., 2016). When calculating moments using BU inverse dynamics, the Ground Reaction Forces (GRF) tend to exert the most significant influence, which means that inertial properties are not as critical in these calculations. Conversely, when determining moments via a TD approach, the moments are primarily determined by inertial properties. Consequently, less precise estimates may be observed when large body segment assumptions are used (Desjardins et al., 1998; Plamondon et al., 1996).

6. Impact of load, height, and asymmetry angle

To our knowledge, the current study represents the first attempt to assess the differences in moment estimates between IMC and OMC systems using biomechanical models within a systematic statistical framework. We analyzed the results in relation to the groups of subjects' load, height, and asymmetry levels, as well as their potential interactions.

Inter-group differences and load had the most pronounced influence on peak moments. In contrast, the influences of height and angle were not significant in our tested scenarios. These findings align with previous research, highlighting that as load mass increases, its effect on low back loading intensifies (Davis et al., 2010; Granata et al., 1999; Lavender et al., 2003; Plamondon et al., 1996; Skals et al., 2021).

Regarding the impact of the variables on the RMSE, the load consistently stood out as the primary factor. Notably, when comparing the IMC system to the gold standard, a Load*Height interaction was observed. Height also significantly differed when comparing the IMC system and the OMC-TD approach. We believe these disparities might arise from the abovementioned variations in segment lengths between the systems.

It is also worth mentioning that the IMU system's sacrum sensor placement was somewhat problematic because no prominent bony landmark can be used for the IMU attachment point. We believe that due to the nature of the lifting and lowering tasks performed, which required forward bending, the sacrum sensor could have moved (mostly upwards) from its original location, affecting the estimated moment results. This challenge has been identified before in the literature (Larsen et al., 2020; Schall et al., 2021). Further research is needed to understand the nature of the variability and the potential options to reduce it, which would make the IMU system promising for field-based risk assessment.

7. Limitations

There are different constraints to consider in this study. First, it assumes that the weight is distributed evenly on both hands, which is improbable because the load's weight distribution may be uneven, particularly if the load has irregular shapes, as with most tools in an industrial work environment. Moreover, this method neglects the potential effects of pushing or pulling loads. Second, the study's participants were all relatively young and healthy, and considering obese participants, a substantial proportion of the working population in the US (Ogden et al., 2014), may lead to more significant soft-tissue artifacts and affect the performance of the IMC system (Bolink et al., 2016). Third, possible imperfections in the scaling of the model may be another source of error that affects the accuracy of the estimates (Konrath et al., 2019). Fourthly, the participants were not restricted in their lifting technique or speed during the lifting and lowering tasks, which could affect the variability of the results (Fleron et al., 2019). Finally, during the study, IMC data was captured on a per-trial basis, with trials lasting around half a minute, so significant errors associated with magnetic disturbance and gyroscopic drift may not have been observed, as the literature has reported such errors in longer trials (Dufour et al., 2021; Robert-Lachaine et al., 2017).

Key points

- The load was the main factor affecting the moment estimate differences between systems in most studied scenarios. Asymmetry (angle), lifting height, and the interactions between variables did not

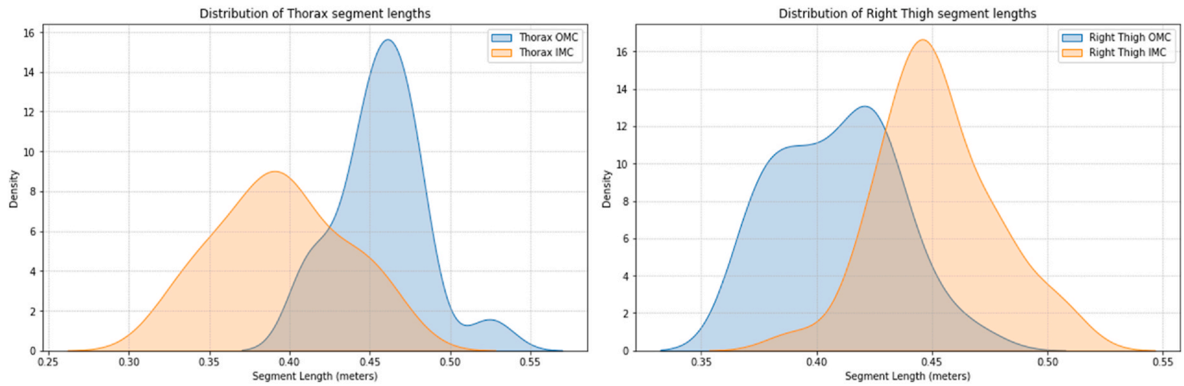

Fig. 10. KDE plots of Thorax segment length (Left) and Right thigh (Right).

offer proper statistical evidence or trends that could invite them to be considered a systematic source of variability.

- When comparing OMC and IMC model estimates for the RMSE, the estimates show close correspondence, which is congruent with the existing literature.
- For the inverse dynamics Top-Down calculations, assigning the weight of the load to the hands is an option that can be explored when the analysis needs to be simplified.
- The IMC-based moment underestimation could be attributed to the significant differences in segment length between biomechanical models.

CRediT authorship contribution statement

Iván Nail-Ulloa: Writing – review & editing, Writing – original draft, Visualization, Validation, Methodology, Investigation, Formal analysis, Conceptualization. **Rong Huangfu:** Methodology, Data curation, Conceptualization. **Michael Zabala:** Data curation, Methodology, Resources, Validation, Writing – review & editing. **Dania Bani Hani:** Data curation. **Nathan Pool:** Data curation. **Howard Chen:** Writing – review & editing, Methodology, Conceptualization. **Mark C. Schall:** Writing – review & editing, Validation, Supervision, Conceptualization. **Richard Sesek:** Resources, Methodology, Conceptualization. **Sean Gallagher:** Writing – review & editing, Supervision, Project administration, Methodology, Investigation, Funding acquisition, Formal analysis.

Declaration of competing interest

The authors declare the following financial interests/personal relationships which may be considered as potential competing interests: Sean Gallagher reports financial support was provided by National Institute for Occupational Safety and Health. If there are other authors, they declare that they have no known competing financial interests or personal relationships that could have appeared to influence the work reported in this paper.

Data availability

Data will be made available on request.

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Appendix 1

Table 1A
Summary of measures of the moment estimates for the different modeling approaches, with values in parenthesis, indicating the percentage difference from the Gold-Standard OMC-BU

Summary measure	Peak Moment			RMSE		
	Peak OMC-BU	Peak IMC-TD	Peak OMC-TD	OMC-BU OMC-TD	OMC-BU IMC-TD	OMC-TD IMC-TD
Lifting (Nm)						
Mean	197.59	171.63 *(-13%)*	208.59 *(+6%)*	16.03	19.07	20.90
SD	45.09	42.50	44.30	6.40	5.94	8.56
Max	320.25	295.35	358.91	39.64	34.74	46.95
Min	110.91	97.66	121.12	4.92	8.20	7.35
25th percentile	161.94	139.39	176.77	11.44	15.28	14.09
50th percentile	195.02	166.87	206.24	15.50	18.05	19.78
75th percentile	231.32	203.23	233.24	19.30	23.23	26.08
90th percentile	258.91	224.61	262.64	25.33	26.86	32.71
Lowering (Nm)						
Mean	188.47	166.24 *(-12%)*	200.57 *(+6%)*	16.95	21.38	21.34
SD	41.10	39.74	40.93	6.79	7.24	8.96
Max	284.01	291.89	329.92	43.30	41.72	47.12

(*continued on next page*)

Table 1A (*continued*)

Summary measure	Peak Moment			RMSE		
	Peak OMC-BU	Peak IMC-TD	Peak OMC-TD	OMC-BU OMC-TD	OMC-BU IMC-TD	OMC-TD IMC-TD
Min	96.55	98.14	120.45	4.22	6.18	6.00
25th percentile	157.62	136.41	171.05	12.28	16.05	15.02
50th percentile	186.59	161.45	199.47	16.06	20.74	19.49
75th percentile	219.73	195.06	224.10	19.64	25.58	26.99
90th percentile	243.39	220.55	249.39	25.09	30.46	33.06

Fig 1A

Bland-Altman plots for Peak Moments for the lifting (Top) and lowering (Bottom), with OMC-BU vs. IMC TD (Left), OMC-BU vs. OMC-TD (Middle), and OMC-TD vs. IMC-TD (Right)

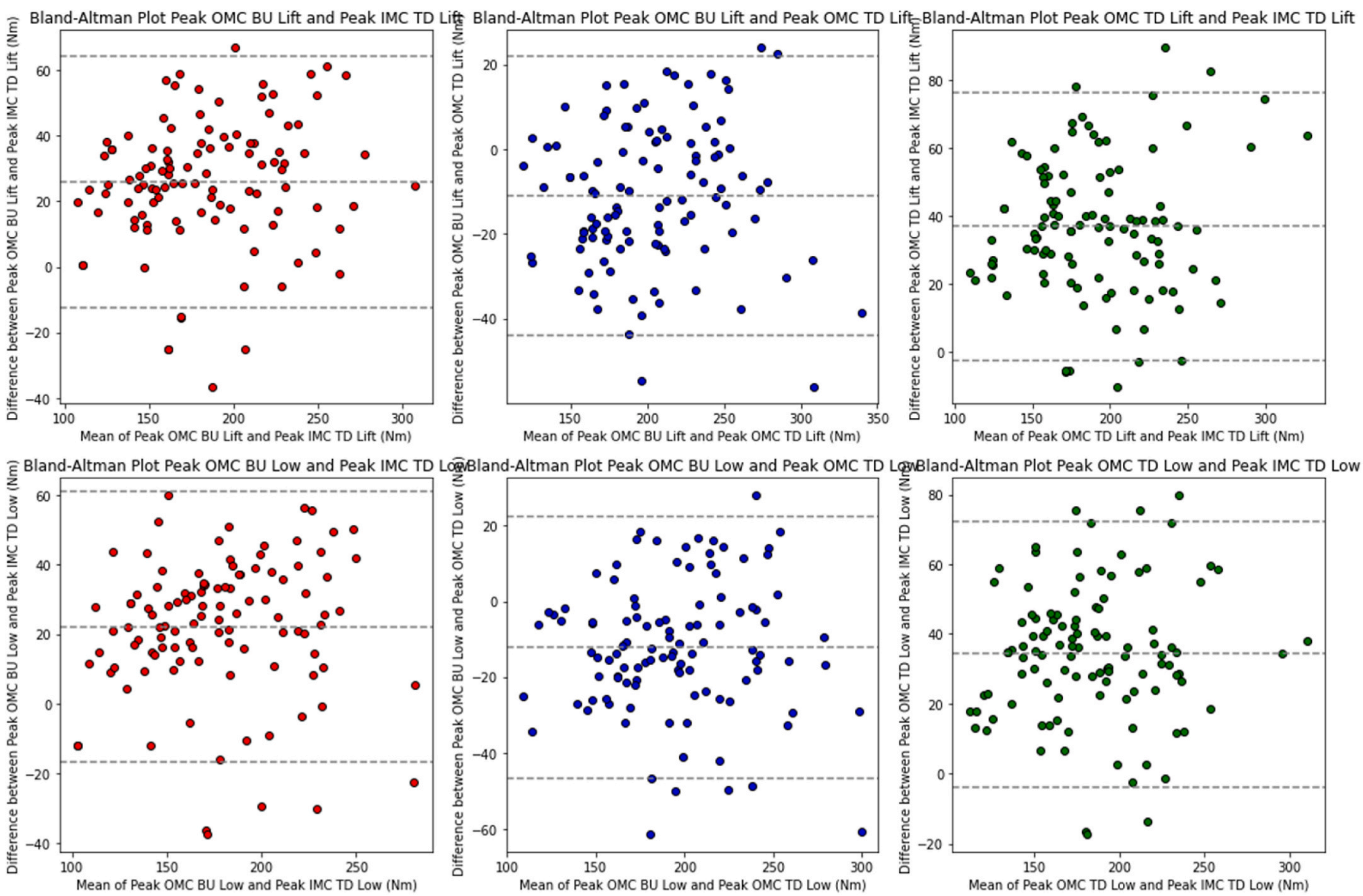

Fig 2A

Scatter plots for Peak Moments for the lifting (Top) and lowering (Bottom), with OMC-BU vs. IMC TD (Left), OMC-BU vs. OMC-TD (Middle), and OMC-TD vs. IMC-TD (Right)

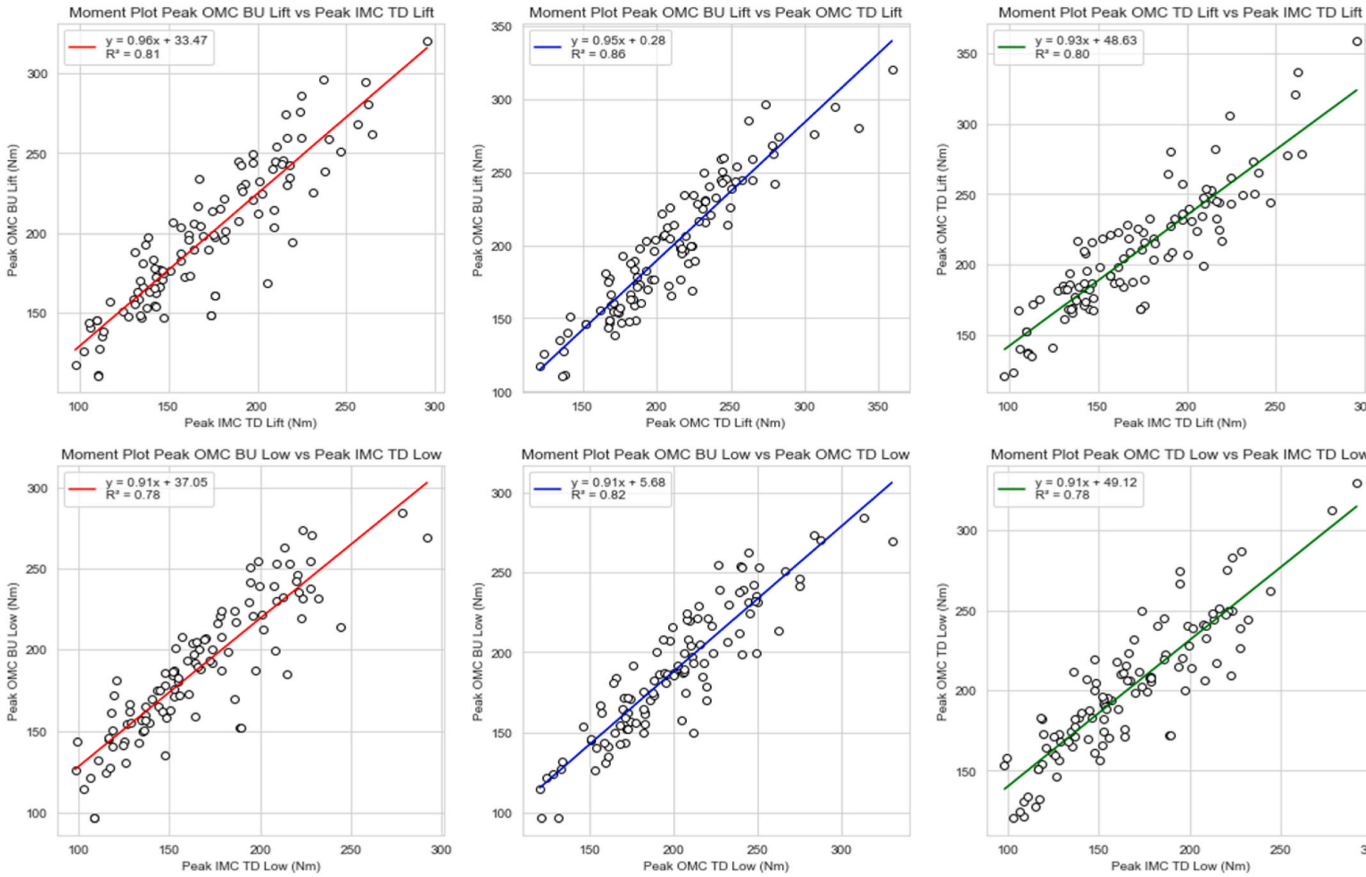

Table 2A

ANOVA results testing the effects of Loads (A), Angle (B), Height (C), and interactions for the RMSE of the moments at L5/S1 during the lifting tasks for the different modeling approaches. Significant effects ($p < 0.05$) are indicated in ***bold/italics.*** Large effect sizes ($\omega^2 > 0.14$) are shown in **bold**.

Factors (df, df Error)	RMSE OMC BU - OMC TD		RMSE OMC BU - IMC TD		RMSE OMC TD - IMC TD	
	p-value	Effect Size ω^2	p-value	Effect Size ω^2	p-value	Effect Size ω^2
Lifting						
Groups (3, 46)	0.462	−0.010	***0.088***	0.103	***0.265***	***0.031***
A - Load (2, 46)	***<0.001***	**0.420**	***<0.001***	**0.671**	***0.019***	**0.145**
B - Angle (2, 46)	0.435	−0.006	0.448	−0.003	0.527	−0.015
C - Height (2, 46)	0.303	0.010	0.072	0.027	0.002	**0.270**
A*B (4, 46)	0.240	0.036	0.613	−0.010	0.345	0.013
A*C (4, 46)	0.255	0.032	***0.003***	0.114	***0.049***	0.140
B*C (4, 46)	0.369	0.008	0.575	−0.008	0.866	−0.060
A*B*C (8, 46)	0.797	−0.073	0.390	0.005	0.844	−0.086
Lowering						
Groups (3, 46)	0.369	0.007	***0.012***	**0.213**	0.293	0.024
A - Load (2, 46)	***<0.001***	**0.529**	***<0.001***	**0.547**	0.587	−0.027
B - Angle (2, 46)	0.209	0.019	0.412	−0.002	0.750	−0.042
C - Height (2, 46)	0.214	0.018	0.416	−0.002	***0.005***	**0.299**
A*B (4, 46)	0.687	−0.026	0.560	−0.011	0.260	0.043
A*C (4, 46)	0.129	0.054	***0.002***	**0.181**	0.119	0.111
B*C (4, 46)	0.385	0.004	0.190	0.028	0.529	−0.023
A*B*C (8, 46)	0.509	−0.010	0.914	−0.056	0.946	−0.156

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