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Short communication

Predicted vs. measured paraspinal muscle activity in adolescent idiopathic scoliosis patients: EMG validation of optimization-based musculoskeletal simulations

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A R T I C L E I N F O

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A B S T R A C T

Musculoskeletal (MSK) models offer great potential for predicting the muscle forces required to inform more detailed simulations of vertebral endplate loading in adolescent idiopathic scoliosis (AIS). In this work, simulations based on static optimization were compared with in vivo measurements in two AIS patients to determine whether computational approaches alone are sufficient for accurate prediction of paraspinal muscle activity during functional activities.

We used biplanar radiographs and marker-based motion capture, ground reaction force, and electromyography (EMG) data from two patients with mild and moderate thoracolumbar AIS (Cobb angles: 21° and 45°, respectively) during standing while holding two weights in front (reference position), walking, running, and object lifting. Using a fully automated approach, 3D spinal shape was extracted from the radiographs. Geometrically personalized OpenSim-based MSK models were created by deforming the spine of pre-scaled full-body models of children/adolescents. Simulations were performed using an experimentally controlled backward approach. Differences between model predictions and EMG measurements of paraspinal muscle activity (both expressed as a percentage of the reference position values) at three different locations around the scoliotic main curve were quantified by root mean square error (RMSE) and cross-correlation (XCorr).

Predicted and measured muscle activity correlated best for mild AIS during object lifting (XCorr's $\geq$ 0.97), with relatively low RMSE values. For moderate AIS as well as the walking and running activities, agreement was lower, with XCorr reaching values of 0.51 and comparably high RMSE values.

This study demonstrates that static optimization alone seems not appropriate for predicting muscle activity in AIS patients, particularly in those with more than mild deformations as well as when performing upright activities such as walking and running.

1. Introduction

Adolescent idiopathic scoliosis (AIS) is a poorly understood, complex three-dimensional deformity of the spine that occurs in the early stages of puberty and affects up to 4 % of the population, with a predominance in females (Cheng et al., 2015). After initial diagnosis, patients are usually treated conservatively with physiotherapeutic approaches such as scoliosis-specific exercises (SSEs). However, aside from the demonstrated positive effects on proprioception, muscle strength and flexibility of the spine, SSEs do not appear to be effective for modulating spinal

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growth in order to minimize curve progression (Day et al., 2019). This is likely at least partially due to a limited understanding of the forces acting on the vertebral endplates during everyday living activities as well as SSEs. One possible approach to better understand these forces is finite element (FE) modeling (Gould et al., 2021). However, most FE models of the AIS spine are loaded with a single generic force representing body weight (see for example Zhang et al. (2021)), while forces exerted by the muscles are largely ignored. Furthermore, the few studies that considered muscle forces only included simulations of static (e.g., upright standing) instead of dynamic activities, and the musculoskeletal (MSK) models used to predict these forces were usually not validated (see for example Kamal et al. (2019)).

For this reason, MSK models of AIS patients should be properly validated before applying them to predict muscle forces for FE simulations. A first step in this direction is to explore whether computational approaches alone realistically predict muscle activity during functional activities that do not involve intentional muscle contractions such as SSEs. This study therefore compares the paraspinal muscle activity around the scoliotic main curve, estimated by static optimization-based MSK simulations of two AIS patients performing different daily activities, with electromyographic (EMG) activity measured in vivo.

2. Methods

2.1. Study participants and data collection

Two patients with mild to moderate right-convex main thoracolumbar AIS ("AIS21": age = 11 years, height = 1.48 m, mass = 32 kg, Cobb angle = 21°, curve apex: T9; "AIS45": age = 15 years, height = 1.56 m, mass = 50 kg, Cobb angle = 45°, curve apex: T8) were recruited and invited for a single visit to the university's movement laboratory. After obtaining written informed consent from their legal guardians, the patients were equipped with 58 *retro*-reflective skin markers according to a previously described configuration (Schmid et al., 2017). In addition, pairs of surface electrodes were placed bilaterally on the erector spinae muscle (approximately 3 cm from the midline) at the level of the apical vertebra, as well as the levels of the vertebrae representing the upper and lower ends of the curve (Cheung et al., 2005) (Fig. 1).

As a reference for the muscle activity measurements, patients were then asked to stand upright for 10 s while holding a 1.5 kg-weight in each hand in front of them (shoulders flexed about 90°). This method resulted on both sides in sufficient muscle activity for adequate normalization. Subsequently, they were asked to perform 5 repetitions of 1) lifting up and putting down a 5 kg-box with a freestyle technique, as well as 2) walking and 3) running on a 10 m-level surface at a self-selected speed.

The three-dimensional positions of the skin markers were recorded using a 16-camera motion capture system (VICON Motion System Ltd, Oxford, UK; sampling frequency: 200 Hz) and two embedded force plates (AMTI Inc., Watertown, MA, USA; sampling frequency: 1 kHz) were used to measure ground reaction forces. Muscle activity was recorded using a telemetric 16-channel surface EMG system (Myon AG, Schwarzenberg, Switzerland; sampling frequency: 1 kHz; in-built hardware filter: 10–500 Hz band-pass). The detailed anatomy of the patients' spinal deformity was obtained using simultaneously captured and spatially calibrated biplanar radiographic images (EOS Imaging, Paris, France), taken within 3 months of the measurements in the movement laboratory.

The local ethics committee granted exemption for this study.

2.2. Patient-specific MSK models

Patient-specific MSK models were created based on previously developed and validated OpenSim-based MSK full-body models for healthy children and adolescents aged 6 to 18 years. These models include a fully articulated thoracolumbar spine (i.e., T1/2 to L5/S1 articulated each with 3 rotational but no translational degrees of freedom) and rib cage as well as age- and gender-specific anthropometrics (i.e., body length and mass distribution, and inertial properties) and muscle strength capacities (Schmid et al., 2020a). The validation included comparisons of model predictions of maximum trunk muscle strength, lumbar disc compressibility, intradiscal pressure, and trunk muscle activity with in vivo studies reported in the literature involving children and adolescents (Schmid et al., 2020a).

Using a fully automated deep neural network-based approach (Galbusera et al., 2019), the position, orientation and height of each vertebra from T1 to L5 was extracted from the biplanar radiographic images by identifying 10 landmarks for each vertebra (8 for the upper and lower endplates, 2 for the pedicle centers) (Fig. 1). Joint centers were thereby determined based on the mean of the intersection points of the two normal vectors of two consecutive endplates with their midplane (Fig. 1). Compared to previous work (Schmid et al., 2020b), where joint centers were defined based on the centroids of the intervertebral space projections, the current method is considered more accurate.

The extracted deformity was then implemented into the pre-scaled MSK full-body model by changing the position and orientation of each vertebra from T1 to L5, and by subsequently re-adjusting the orientations of the ribs (but not their geometry), lumped head-neck body, arms, and segmental centers of mass (CoM) to match those of the undeformed model. In addition, since the deformity implementation process resulted in an inappropriate displacement of the modeled markers, they were re-adjusted based on the optimal position (most dorsal tip of spinous processes) determined manually from the radiographic images to reflect the positions of the markers placed in the movement laboratory.

To reduce the computational effort, all rib joints were locked (which did not affect the movement of the spinal joints), and the intercostal

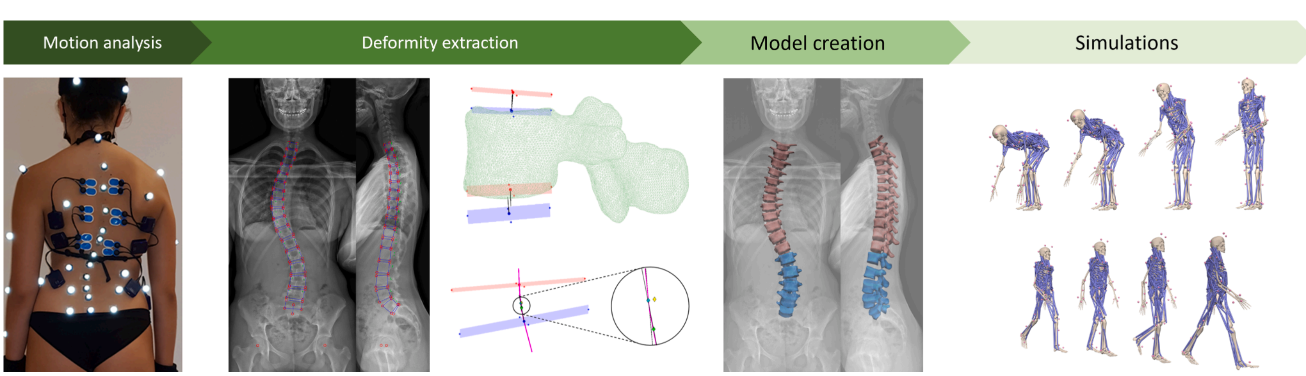

Fig. 1. Personalized OpenSim-based musculoskeletal modeling workflow for patients with AIS, including full-body motion analysis, model creation from biplanar radiographs, and simulations of muscle forces during various functional activities using an experimentally controlled backward approach (i.e., inverse kinematics and static optimization).

muscles were removed.

2.3. Simulation of functional activities

A backward simulation approach was used to predict paraspinal muscle forces during each functional activity (Fig. 1). Skin marker data were pre-processed using the Nexus software (VICON Motion System Ltd, Oxford, UK) to perform the reconstruction, labeling, filtering of the markers, as well as the manual identification of the temporal events that determine the different walking and running cycles. The lifting up phase of the object lifting task was automatically identified using a custom MATLAB routine (MathWorks, Inc., Natick, MA, USA) (Schmid et al., 2022).

Joint angles were calculated using the OpenSim Inverse Kinematics tool, without applying any kinematic constraints such as pre-defined intervertebral motion ratios (Alemi et al., 2021). To estimate individual muscle activity, we used the OpenSim Static Optimization tool with a cost function to minimize the sum of squared muscle activations (Herzog, 1987). To compensate for the lack of modeled intra-abdominal pressure as well as passive stiffness resulting from joint capsules, ligaments and fasciae (e.g., thoracolumbar fascia), muscle strength capacities were increased by a factor of 1.5, and the intervertebral joints were actuated in all directions using coordinate actuators with optimal torques (i.e., torques generated with an activation of 1.0) set at 3 Nm for T1/T2-T11/T12, 3.75 Nm for T12/L1-L2/L3, and 10 Nm for L3/L4-L5/S1. These values were initially determined using data from the literature, and subsequently adjusted based on the optimization performance of multiple preliminary simulations. We thereby ensured that they were kept as low as possible to make sure that the optimizer preferred the muscles over the actuators to generate the required torques. The weights for the reference position were modeled by applying two vertically directed static forces to the hands, while the box for the lifting task was modeled by modifying the mass, CoM location, and inertial properties of the hands to match those of a combined hands-box body, therefore taking into account the inertial forces of lifted weight.

2.4. Validation of model predictions

The predicted paraspinal muscle activity was validated by comparing it to the paraspinal muscle activity measured in vivo at the three different locations (apex, upper and lower ends) on both the convex and concave sides of the scoliotic main curve. EMG signals were therefore filtered using a digital 30 Hz high-pass Butterworth filter to remove electrocardiogram contamination (Drake and Callaghan, 2006), and then processed by creating an envelope using rectification and a moving average with a window of 100 ms. For the model predictions, the activity of the muscle fibers whose paths crossed the electrode locations were averaged to determine the model-estimated muscle activations for each time frame (Alemi et al., 2023). To allow comparisons between predicted and measured muscle activity, all functional activity data were time-normalized to 101 data points and expressed as a percentage of the reference position values. We evaluated the following parameters: predicted and measured muscle activity on both the convex and concave sides of the main scoliotic curve, as well as the ratios between muscle activity on the convex and concave sides at the apex and the upper and lower ends of the curve. Comparisons were quantified using root mean square errors (RMSE) and cross-correlation coefficients (XCorr) (Wren et al., 2006).

3. Results

Predicted and measured muscle activity as well as ratios during object lifting correlated very well at all three spinal levels (XCorr $\geq$ 0.97 for AIS21, and $\geq$ 0.93 for AIS45), with RMSE's for ratio of $\leq$ 0.5 for AIS21 and $\leq$ 0.7 for AIS45 (Fig. 2, and Figs. A1 and A2 in the electronic supplementary material). However, while the RMSE's for muscle activity were $\leq$ 70 % for AIS21, they reached values of almost 300 % on the convex and 120 % on the concave side for AIS45. During walking, the XCorr's ranged from 0.71 to 0.95 for muscle activity and from 0.60 to 0.68 for ratio, with RMSE's of up to 85.7 % for muscle activity and 54.3 for ratio. During running, the XCorr's ranged from 0.42 to 0.94 for muscle activity and from 0.51 to 0.78 for ratio, with RMSE's of up to 170.4 % for muscle activity and 110.5 for ratio.

4. Discussion

This study aimed at validating personalized experimentally controlled MSK models for predicting paraspinal muscle activity during functional activities using a static optimization approach. The results indicated that the optimization-based predictions agreed best with the in vivo-measured muscle activity for the patient with less scoliotic deformation (AIS21) during object lifting. However, the agreement decreased with more scoliotic deformation (AIS45) as well as more dynamic and upright activities (i.e., walking and running).

The excessively high convex-to-concave ratios found for the walking and running tasks can likely be attributed to the fact that the applied static optimization approach did not adequately account for physiological or pathological muscle co-contractions, which often resulted in very low or even completely absent muscle forces on the concave side of the curve. Static optimization determines the muscle forces required to achieve equilibrium of moments in each intervertebral joint (i.e., a net joint moment of zero), whereby antagonistic muscle activity is largely underestimated or even lacking (Kian et al., 2019). Hence, to prevent the spine from collapsing towards the concave side during upright standing, only the muscles on the convex side needed to be activated.

To overcome this problem, EMG-assisted optimization (EMG-AO) can be used. Banks et al. (2022) recently developed a MATLAB-based framework for incorporating EMG-AO in OpenSim, and used it to study the biomechanics of the lumbar spine during gait. Compared to static optimization alone, the EMG-AO approach predicted higher lumbar joint loads and muscle activations that better matched the EMG patterns of the individual participants (Banks et al., 2022). Future simulations of trunk muscle activity in patients with AIS should therefore incorporate EMG-AO.

Even though the models were personalized for body height/segmental lengths and spinal shape, several patient-specific tissue properties were missing. First, the model did not include any passive stiffness properties, which would be necessary to account for the torques generated by joint capsules, ligaments, and fasciae. To compensate for this, we increased muscle strength capacity and used artificial torque generators, which, however, might likely not have reflected the true contribution of soft tissues. In fact, the actuator-based contribution (particularly in the lower lumbar segments) was often higher than the generally accepted 5 % of the external joint moment (Hicks et al., 2015), indicating that the already stronger muscles were still not able to provide at least 95 % of required joint torque. In addition, it cannot be excluded that increasing the maximum force capacity of the muscles affected the computed muscle activation pattern. Future work should therefore not only focus on the implementation of passive stiffness properties but also consider enhancing the model with additional spinal muscles and even modeling the thoracolumbar fascia to enable sufficient torque generation for simulating functional activities in complex spinal deformities.

Furthermore, our EMG normalization approach might have resulted in larger error proneness for the muscles with lower activity during the reference position. Even though the spinal deformity hinders symmetric bilateral muscle contractions, this error proneness might be reduced by implementing a more sophisticated normalization protocol such as a combination of different positions and motions.

Finally, the model lacked adequate information on patient-specific segmental mass distribution (i.e., the location of the CoM for each trunk segment) and muscle geometry. This could have had an important

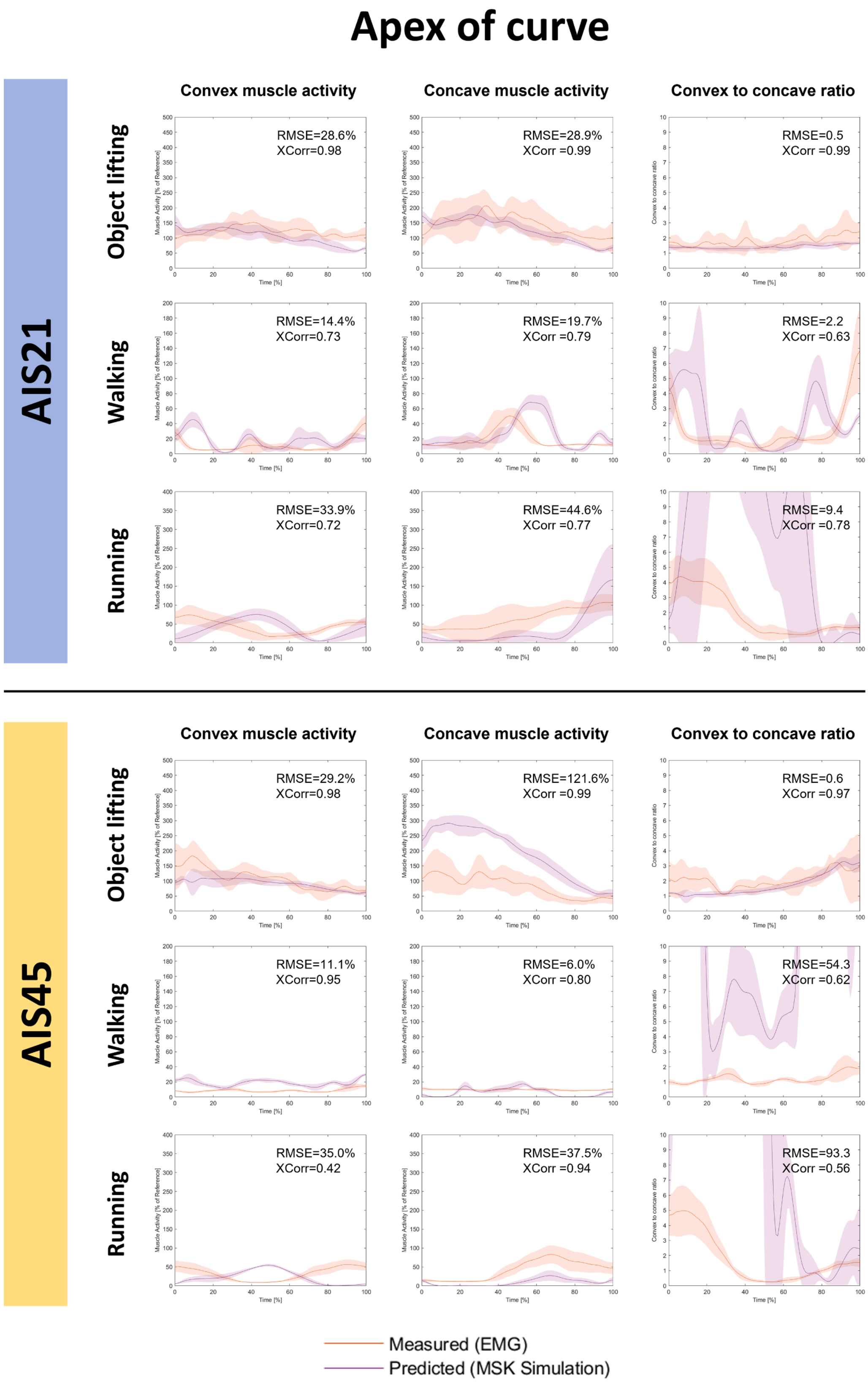

Fig. 2. In vivo measured EMG activity and predicted muscle activity at the height of the scoliotic curve apex in the patients AIS21 and AIS45 during object lifting, walking, and running. Left and middle column: convex and concave activity in relation to the reference position. Right column: ratio between the convex and concave side activity. RMSE = root mean square error; XCorr = cross-correlation coefficient.

impact on the external moments and the optimization-based prediction of muscle forces. To avoid these shortcomings, CoM positions and muscle cross-sectional areas could be derived from CT or MRI scans (Anderson et al., 2012; Keenan et al., 2014).

In conclusion, this study demonstrates that static optimization alone seems not appropriate to predict muscle activity in patients with AIS, particularly in those with more advanced scoliotic deformations (i.e., moderate AIS or more) and when simulating more dynamic and upright activities such as walking and running. Future models of AIS patients should therefore incorporate passive stiffness properties, segmental mass distribution and muscle geometry, and might even be enhanced with additional force generating elements. Moreover, simulations should be performed using an EMG-assisted optimization approach.

CRediT authorship contribution statement

Cedric Rauber: Writing – original draft, Methodology, Investigation, Formal analysis, Conceptualization. **Dominique Lüscher:** Writing – review & editing, Methodology, Conceptualization. **Lucile Poux:** Writing – review & editing, Investigation, Data curation, Conceptualization. **Maria Schori:** Writing – review & editing, Resources, Investigation, Conceptualization. **Moritz C. Deml:** Writing – review & editing, Resources, Conceptualization. **Carol-Claudius Hasler:** Writing – review & editing, Resources, Conceptualization. **Tito Bassani:** Writing – review & editing, Methodology. **Fabio Galbusera:** Writing – review & editing, Software, Methodology. **Philippe Büchler:** Writing – review & editing, Supervision, Methodology, Funding acquisition, Conceptualization. **Stefan Schmid:** Conceptualization, Funding acquisition, Data curation, Writing - original draft, Visualization, Methodology, Supervision, Resources, Project administration.

Declaration of competing interest

The authors declare that they have no known competing financial interests or personal relationships that could have appeared to influence the work reported in this paper.

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Appendix A. Supplementary material

Supplementary data to this article can be found online at https://doi.org/10.1016/j.jbiomech.2023.111922.

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