



Evaluation of EMG-Assisted and EMG-Driven control modes in patients with medial compartment knee osteoarthritis

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ABSTRACT

Knee osteoarthritis (OA) is associated with higher-than-normal knee joint contact forces (KJCFs) during walking which cannot be easily measured. KJCFs can be estimated using neuromusculoskeletal (NMSK) modelling in electromyogram (EMG)-driven and –assisted control modes. Previous research has not examined which control mode is most appropriate for estimating KJCFs in patients with knee OA. This study aimed to evaluate a NMSK modelling framework using both control modes in patients with medial-dominant knee OA. First, EMG-assisted mode was hypothesized to better track ID-computed joint moments. Second, KJCFs estimated using the two control modes were hypothesized to differ.

Gait data were measured from 27 patients with medial-dominant tibiofemoral knee OA. An OpenSim model was scaled to patient-specific anthropometrics. Inverse kinematics, inverse dynamics, and muscle analysis were performed. Resulting joint angles, moments, musculotendon kinematics, and muscle activations were input into the Calibrated Electromyography Informed Neuromusculoskeletal Modelling Toolbox. Muscle forces and KJCFs were estimated using EMG-driven and –assisted control modes.

EMG-assisted mode better tracked knee flexion moments ($RMSE = 2.7 \pm 2.1Nm$, $R^2 = 0.9 \pm 0.1$) and estimated a higher, albeit non-significant, second peak medial compartment KJCF ($2.3 \pm 1.1BW$) compared to EMG-driven mode ($RMSE = 12.6 \pm 3.9Nm$, $R^2 = 0.6 \pm 0.2$, $KJCF = 2.1 \pm 0.9BW$). EMG-assisted control mode may therefore be more appropriate for evaluating KJCFs in patients with knee OA.

1. Introduction

Knee osteoarthritis (OA) has a complex pathogenesis associated with higher-than-normal forces across the tibiofemoral joint (Andriacchi and Mündermann, 2006). Although there are no proven disease modifying interventions, many conservative and surgical interventions aim to improve symptoms and function by altering biomechanics (Hall et al., 2019; Madry, 2022; Marriott and Birmingham, 2023; S. Starkey et al., 2022). These interventions aim to lower medial knee joint contact forces (KJCFs) that are associated with progression of medial-dominant knee OA (Andriacchi, 1994; Lewek et al., 2004). KJCFs cannot be measured

directly without instrumented knee implants (such as in Halder et al., 2012). Knee adduction moments (KAMs) calculated using inverse dynamics (ID) are commonly used as proxy measures for KJCFs as they cannot be directly measured within the intact knee (Holder et al., 2023; Kutzner et al., 2013; Miyazaki et al., 2002; Walter et al., 2010; Zhao et al., 2007). However, KAMs correlate poorly with articular loads as they are estimated from external load contributions to KJCFs using experimental kinematic and kinetic data only (Saxby et al., 2016).

Neuromusculoskeletal (NMSK) modelling approaches can accurately estimate articular forces as they consider internal muscle forces derived from electromyogram (EMG) data when calculating KJCFs (Bennett

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Table 1
Patient Demographics, Clinical Characteristics, and Gait Speed.

	Values (Mean $\pm$ SD)*
Patient Demographics	
Age (years)	53 ($\pm$ 5)
Sex (male/female, # of patients)	(20/9)
BMI (kg/m ²)***	29.8 ($\pm$ 4.2)
MAA**	-6.5 ($\pm$ 1.8)
Clinical Characteristics	
KL Scale Grade (N)****	1(4); 2(13); 3(12); 4(0)
Gait Speed	
m/s	1.08 ($\pm$ 0.16)

* Values are the mean varus angle $\pm$ SD unless otherwise stated.

** MAA = mechanical axis angle. Negative values indicate varus.

*** BMI = body mass index.

**** Kellgren & Lawrence (KL) scale grade of osteoarthritis severity.

et al., 2022). NMSK models can be executed using EMG-driven and –assisted control modes. EMG-driven mode uses recorded EMG and kinematic data to compute muscle forces. This open-loop control mode neglects ID-computed joint torques (Sartori et al., 2014) and the net actions of these computed muscle forces across the joint can be inconsistent with ID-computed joint moments. These inconsistencies likely stem from errors in experimental EMG measurement or normalization, or assumptions regarding the activity levels of muscles that were not measured.

Alternatively, EMG-assisted mode optimizes predicted muscle forces to minimize an objective function that considers both muscle activation and ID-computed joint moment errors (Sartori et al., 2014). As a result, EMG-assisted mode is less sensitive to errors in experimental EMG data and thus is assumed more appropriate for studies of pathological populations, such as patients with knee OA, for whom high-quality EMG data can be difficult to acquire (Gerus et al., 2013; Sartori et al., 2014). Both EMG-driven and –assisted control modes have been employed in studies involving healthy populations and patients with knee OA (Gerus et al., 2013; Hall et al., 2019; Saxby et al., 2016; S. Starkey et al., 2022). Previous research, however, has not examined which control mode is most appropriate for estimating KJCFs in patients with knee OA.

This study aimed to evaluate a patient-specific NMSK modelling framework using two NMSK control modes in patients with knee OA. First, EMG-driven and –assisted control mode-computed knee flexion moments (KFMs) were compared against ID-computed KFMs to assess joint moment tracking accuracy. Given EMG-assisted mode allows optimization to adjust muscle excitations and improve joint moment tracking accuracy, it was hypothesized that EMG-assisted control mode would better track the ID-computed joint moments (i.e., higher coefficient of determination (R^2) and lower root mean squared error (RMSE)). Second, KJCFs computed using EMG-driven and –assisted control modes were compared across stance using Statistical Parametric Mapping (SPM). First and second peaks were also computed to allow for comparison with previously published studies. It was hypothesized that KJCFs estimated using the two control modes would differ, highlighting the importance of control mode selection in patients with knee OA.

2. Methodology

2.1. Participants

Twenty-seven patients (Table 1) with symptomatic and radiographic medial-dominant tibiofemoral knee OA participated in this study. Participants were recruited as a subgroup of patients from a larger study at the Fowler Kennedy Sport Medicine Clinic, London, Ontario Canada (Primeau et al., 2020). Recruitment occurred after approval from the Research Ethics Board for Health Sciences Research Involving Human Subjects of the University of Western Ontario was obtained. Standing radiographs were taken by a radiology technician following standard

procedures (Specogna et al., 2007). Eligible patients had neutral to varus alignment and greater joint space narrowing in the medial tibiofemoral compartment. Patients with valgus alignment, joint replacement or high tibial osteotomy on either limb, multi-ligamentous instability, inflammatory or infectious arthritis of the knee, neurologic conditions affecting gait, or possible pregnancy were excluded.

2.2. Gait data collection

Lower extremity gait biomechanics were measured at the Wolf Orthopedic Biomechanics Laboratory at Western University. Participants walked barefoot on level ground at their self-selected pace while EMG, kinematic, and kinetic data from the study limb were recorded. A minimum of 3 and maximum of 5 clean trials were collected, ensuring the foot of the study limb landed in the middle of the force plate. Kinematic data were collected using a 10-camera motion capture system (Motion Analysis Corporation, Santa Rosa, CA, USA) and a full-body passive reflective modified Helen Hayes marker set, sampled at 60 Hz (Moyer et al., 2013). Kinetic data were collected using a floor-embedded force plate (AMTI, Watertown, MA, USA) and sampled at 600 Hz. Surface EMG was collected using electrodes placed over the muscle bellies of the rectus femoris (RF), vastus medialis (VM), vastus lateralis (VL), medial hamstring (MH), lateral hamstring (LH), medial gastrocnemius (MG), lateral gastrocnemius (LG), and tibialis anterior (TA) on the limb that was being studied (DeLuca, 1997). Data were acquired across 8 channels at 600 Hz using Ag surface electrode pairs arranged in a bipolar configuration spaced 10 mm apart (Trigno, Delsys, Natick, MA). This system has an input impedance of 10 G Ω , a common mode rejection ratio of 80 dB at 60 Hz (De Luca, 1997), and a bandwidth of 20–450 Hz. EMG data were sampled at 600 Hz as differences between results computed using our musculoskeletal modelling framework with EMG data collected at 1200 Hz and reanalyzed with EMG data downsampled to 600 Hz demonstrated differences less than 1 N in the estimated KJCFs (based on four participants). After the gait trials, participants completed four exercises to elicit their maximum voluntary isometric contractions (MVIC) for each muscle group (Rueterbories et al., 2010). MVIC exercises that are detailed in S2 were held for a minimum of 3 s and verbal encouragement was given. MVIC exercises were completed after the walking trial to ensure they did not affect the participants' muscle activity patterns while walking.

2.3. Data processing

Gait data were processed within MATLAB (The MathWorks, Inc. Natick MA) using the Motion Data Elaboration Toolbox for Musculoskeletal Applications (MOtoNMS) toolbox (Mantoan et al., 2015). Kinematic and kinetic data for all static and gait trials were lowpass filtered using a zero-lag 2nd order Butterworth filter with a cut-off frequency of 6 Hz (Mantoan et al., 2015). Knee and ankle joint centers were obtained by taking the midpoint between the respective medial and lateral knee and ankle markers during the static trial. The hip joint center was determined by a sphere fit of the center of the knee relative to the pelvis during a leg swing trial using a custom MATLAB script.

EMG data were highpass filtered (20 Hz lower cut-off), full wave rectified, and lowpass filtered using a zero-lag 2nd order Butterworth filter with a 6 Hz cut-off frequency to calculate linear envelopes (De Luca et al., 2010). Linear envelopes were amplitude-normalized to the maximum value of all recorded linear envelopes (considering MVIC and gait trials) for each muscle (Lloyd and Besier, 2003). All kinematic, kinetic, and EMG data from each trial were truncated to include only the stance phase of gait (Rueterbories et al., 2010), with a margin of 0.33 s added to the beginning and end of each trial required for running the Calibrated Electromyography Informed Neuromusculoskeletal Modelling Toolbox (CEINMS). Electromyography data was visually screened at this stage of processing by a single investigator to ensure they met requirements for neuromusculoskeletal modelling (Akhundov et al.,

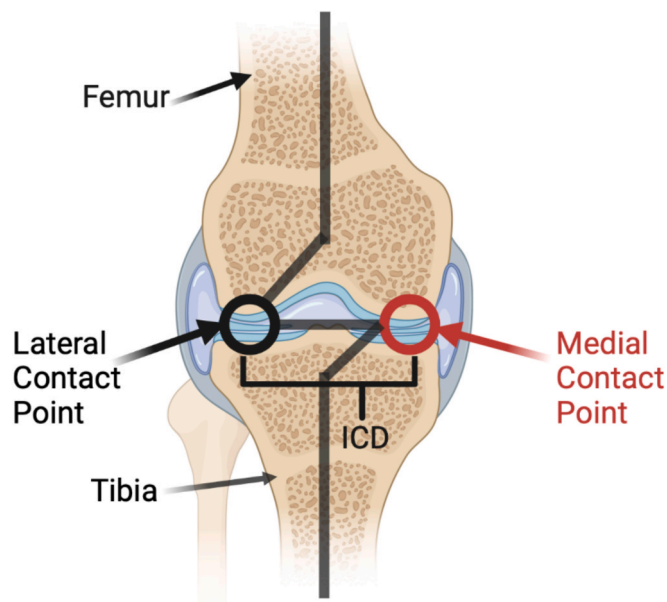


Fig. 1. Adaptation of the Saxby et al. (2016) tibiofemoral mechanism. Two contact points are modelled as hinges. The two contact points allow joint forces and kinematics to be calculated about the medial (red circle) and lateral (black circle) compartment of the tibiofemoral joint. The two contact points are separated by the participant's patient-specific ICD (black bracket) obtained via hip to ankle radiograph. The segment corresponding to the medial contact point outlined in red was also modified to predict the participant's frontal plane alignment angle during scaling (Bowd et al., 2023). Created with [biorender.com](#). (For interpretation of the references to colour in this figure legend, the reader is referred to the web version of this article.)

2019).

2.4. Musculoskeletal modelling

2.4.1. Modifications to base model

Gait biomechanics were calculated in OpenSim v4.1 (Delp et al., 2007) using a generic MSK model (gait2392) originally developed for a study on running dynamics (Hamner et al., 2010). The one-degree-of-freedom (DOF) knee model was modified to compute three-dimensional knee moments and tibiofemoral contact forces (Saxby et al., 2016). Modifications included adding an internal/external rotational DOF to the knee (15° and 5° , respectively), as described by Saxby et al. (2016). Additionally, knee abduction/adduction DOFs were added to calculate net frontal plane joint moments and muscle–tendon moment arms with respect to the medial and lateral compartments of the knee (Fig. 1). Although the angle at the knee abduction/adduction DOF can be adjusted when scaling (by up to 5° adduction and 10° abduction), they remained locked during motion analyses, consistent with previous studies (Saxby et al., 2016). The first contact point (the medial contact point) was located on the medial femoral condyle to calculate medial compartment KJCFs. The second contact point (the lateral contact point) was located on the lateral portion of the tibial plateau to calculate lateral compartment KJCFs. The location of each contact point on the tibial plateau of the generic model was adjusted to match the distance from the patient-specific medial and lateral femoral contact points from the joint center. Locations were obtained as the distance from the most prominent point of the femoral condyles to the knee joint center in each patient's radiograph.

2.4.2. Model scaling

The medial contact point was permitted to match the static pose data, accounting for the patient-specific frontal plane mechanical axis angle. This step was performed to account for the large knee varus angle

within the inclusion criteria for this current study (Bowd et al., 2023). It was also assumed that the deformity primarily affected the tibia segment (Bowd et al., 2023). Each model was linearly scaled with the torso, pelvis, and foot segments scaled to the participant's height, and the thigh and shank segments linearly scaled to match the participant's marker positions in the static trial. After scaling, the medial contact point was locked to the model-predicted mechanical axis angle (mean of $7.6^\circ \pm 2.7^\circ$ varus).

2.4.3. Motion analysis

Gait marker trajectories and kinetic data were input into each scaled model to perform inverse kinematics (all markers were given a weight of 10), ID, and muscle–tendon unit (MTU) kinematics analyses (Saxby et al., 2016). Following OpenSim best practices (Hicks et al., 2015), marker errors were reviewed, and 2 participants were excluded due to exceeding the recommended marker error threshold of < 2 cm and RMSE < 1 cm.

2.4.4. Neuromusculoskeletal modelling

OpenSim-computed joint angles, sagittal plane moments, musculotendon kinematics, and muscle activations were provided as inputs to CEINMS for simulations using both the EMG-driven and the EMG-assisted control modes (Pizzolato et al., 2015). Eight lower extremity EMG linear envelopes were mapped to 11 MTU excitations for both control modes, as previously described by Lloyd and Besier (2003). The remaining lower extremity muscle activations, corresponding to 32 MTUs, were synthesized to improve the prediction of muscle forces spanning the hip and ankle joints. These additional MTUs were not included with the EMG-driven mode. Using the technique described by Gerus et al. (2013), activation dynamics and muscular model parameters were calibrated to minimize a cost function based on hip, knee, and ankle sagittal plane joint moments. Models were calibrated using input data from a single trial (the first gait trial) for each participant with weighting coefficients adjusted based on the error between model inputs to outputs.

The subject-specific NMSK models were executed in CEINMS using both the EMG-driven and the EMG-assisted control modes for the remaining 2–4 gait trials that were not used during calibration. For the EMG-driven mode, 11 recorded activations and kinematic data were used to compute muscle forces by deriving neural excitations from the activations to model the muscle's twitch response (Ceseracciu and Reggiani, 2015; Pizzolato et al., 2015). The MTU kinematics and muscle activations were then used as an input for a Hill-type muscle activation model (Pizzolato et al., 2015).

The EMG-assisted control mode was driven by all 43 MTUs using an objective function that included three components α , β , and γ described in detail by Sartori et al. (2014). Within our study, the α coefficient was held constant ($\alpha = 1$), following CEINMS guidelines (Ceseracciu and Reggiani, 2015). The β and γ coefficients were set to 1.5, based on an analysis that manually varied coefficient values greater than 1 to find the lowest values of the weighted sum across $n = 5$ participants; a method that is described in detail by Sartori et al. (2014). CEINMS-computed outputs for each NMSK control mode included adjusted muscle activations and hip, knee, and ankle sagittal plane moments, and estimated muscle forces. Data were cropped to the stance phase only (removing the 0.33 s margin previously included).

2.4.5. Joint contact force analysis

CEINMS-computed forces for muscles spanning the knee (11 for EMG-driven mode, 13 for EMG-assisted mode) were provided as inputs to an established algorithm for computing medial and lateral KJCFs (Winby et al., 2009). For each participant, the total knee contact force was calculated by summing the medial and lateral compartment contact forces. Resulting KJCFs were time normalized to 100 points across the stance phase and amplitude normalized to the participant's body weight (BW).

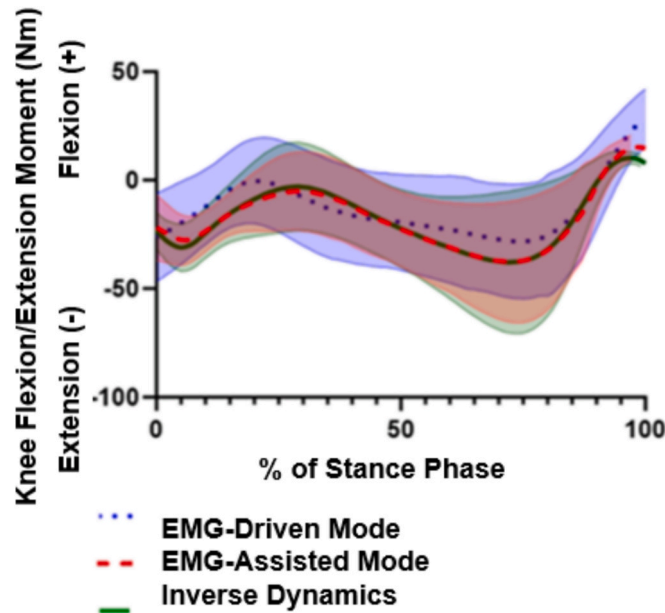


Fig. 2. Ensemble average (± 1 standard deviation) knee flexion/extension moments computed using EMG-driven (blue) and –assisted (red) neuromusculoskeletal modelling control modes and flexion/extension moments from inverse dynamics (ID) (green) across the stance phase of gait for $n = 29$ participants with knee OA. (For interpretation of the references to colour in this figure legend, the reader is referred to the web version of this article.)

The first and second peaks of the total and medial compartment KJCFs were recorded by extracting the highest value in the first and second half of the stance phase, respectively. For the lateral compartment KJCF, the first peak was extracted as the highest value from the first 20 % of the stance phase and the second peak was extracted as the highest value between 50 to 90 % of the stance phase. These portions of the stance phase were extracted based on a previous study examining the effects of a surgical intervention on patients with knee OA that identified statistically significant differences in lateral compartment KJCFs within these ranges of the stance phase (De Pieri et al., 2023).

2.5. Data analysis

To quantify model tracking accuracy, R^2 and RMSE were calculated by comparing KFM computed with EMG-driven and –assisted control modes to ID-computed KFM (Hoang et al., 2019; Pizzolato et al., 2015). R^2 and RMSE were compared between control modes using a paired-sample t -test. Total KJCF calculated with EMG-driven and –assisted control modes were compared across the stance phase using a paired samples t -test via SPM (Pataky 2010). KJCF peaks extracted from the stance phase (total, lateral compartment, medial compartment) were also compared between control modes using paired samples t -tests. All statistical analyses were performed using MATLAB with significance set at 0.05.

3. Results

3.1. Model tracking

Knee flexion moment tracking accuracy was significantly better ($p = 0.001$) for EMG-assisted (RMSE = 2.7 ± 2.1 Nm, $R^2 = 0.9 \pm 0.1$ across stance) compared to EMG-driven (RMSE = 12.6 ± 3.9 Nm, $R^2 = 0.6 \pm 0.2$ across stance) control mode (Fig. 2).

Table 2

Descriptive Statistics and Dependent Samples t -Test Results for Total, Medial, and Lateral Compartment Knee Joint Contact Forces for EMG-Driven and EMG-Assisted NMSK Control Modes. Peak forces measured in TKA patients ($n = 5$) using instrumented implants are shown for comparison (Halder et al., 2012).

		Halder et al. (2012) BW	EMG-Driven BW ($\pm$ SD)	EMG-Assisted BW ($\pm$ SD)	EMG-Driven vs EMG-Assisted p -value
Total Contact Force	First Peak	2.2	2.7 (± 0.8)	2.6 (± 0.8)	0.53
	Second Peak	2.7	2.7 (± 0.9)	2.9 (± 1.2)	0.55
	Peak				
Medial Compartment Contact Force	First Peak	1.5	2.2 (± 0.8)	2.1 (± 0.7)	0.26
	Second Peak	1.8	2.1 (± 0.9)	2.3 (± 1.1)	0.06
	Peak				
Lateral Compartment Contact Force	First Peak	0.6	0.8 (± 0.4)	0.9 (± 0.5)	0.08
	Second Peak	0.9	0.9 (± 0.5)	1.0 (± 0.6)	0.35
	Peak				

Values are presented as mean ($\pm$ SD), p -value.

3.2. Knee joint contact forces

First and second peak KJCFs were not statistically different between control modes (total, medial, or lateral compartment) (Table 2). The largest difference occurred in the medial compartment KJCF where EMG-assisted mode calculated a greater second peak contact force (2.3 ± 1.1 BW) than EMG-driven mode (2.1 ± 0.9 BW). Total KJCF was lower in the first 10 % of stance when calculated using EMG-driven vs EMG-assisted mode. Also shown in Table 2, computed KJCFs compared favorably with those measured using instrumented knee implants (Halder et al., 2012).

4. Discussion

This study aimed to evaluate a patient-specific NMSK modelling framework using two NMSK control modes in patients with knee OA. When first evaluating model tracking accuracy, the EMG-assisted control mode demonstrated significantly lower RMSE and higher R^2 than the EMG-driven control mode. Consistent with our hypothesis and in line with Sartori et al. (2014), better tracking was expected with EMG-assisted mode as it allows optimization to adjust muscle excitations and improve joint moment tracking accuracy. Our EMG-driven joint moment tracking errors (RMSE = 12.6 ± 3.9 Nm) were consistent with previous reports (10–13 Nm) using a similar model in patients with knee OA (Gerus et al., 2013). Our EMG-assisted joint moment tracking errors (RMSE = 2.7 ± 2.1 Nm) were smaller than previous reports using a similar model in patients with hip OA (5.4 Nm) and a healthy population (7.0 Nm) indicating greater joint moment tracking accuracy within our results (Hoang et al., 2019; Pizzolato et al., 2015). In addition, our EMG-assisted results showed comparable R^2 values to existing literature (Hoang et al., 2019; Sartori et al., 2014) to further indicate a strong correlation and a high consistency between our ID-computed and NMSK model-computed KFM.

There were no statistically significant differences for the first or second peak of the total medial compartment, or lateral compartment KJCFs. SPM results indicated statistically significant differences within the total, medial, and lateral KJCF waveforms during the first 10 % of the stance phase. These differences reflect significant waveform variations between control modes, that were not captured by peak-based analysis alone. Furthermore, results suggest that while both modes yielded similar first and second peak KJCF predictions, the temporal patterns were influenced by the control mode.

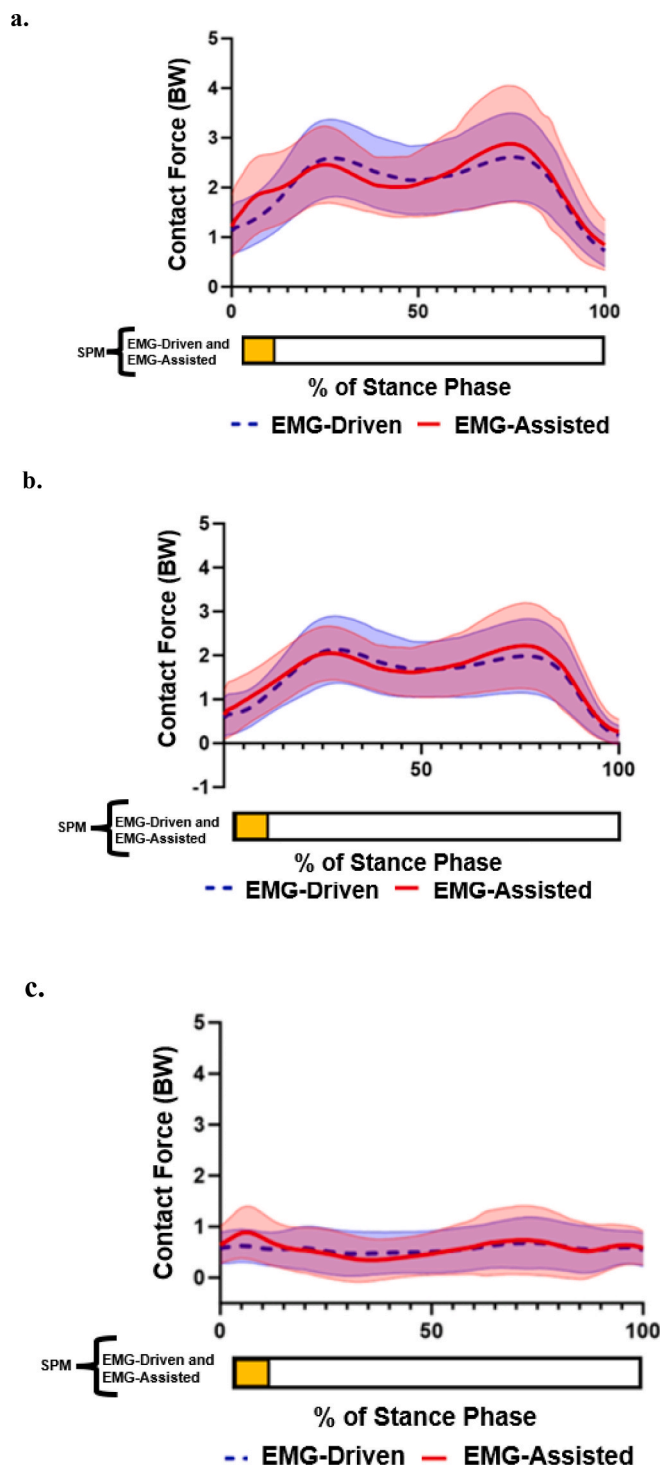


Fig. 3. Ensemble average (± 1 standard deviation) waveforms and statistical parametric mapping for a. Total contact force, b. Medial compartment contact force, and c. Lateral compartment contact force computed by EMG-driven (blue) and –assisted (red) neuromusculoskeletal (NMSK) modelling control modes. (For interpretation of the references to colour in this figure legend, the reader is referred to the web version of this article.)

This study validated a NMSK modelling framework using EMG-assisted and EMG-driven control modes to assess the comparability of their estimated KFMs and KJCFs to ID and the literature. In addition, our findings provide insight into how different control modes influence joint moment predictions and highlight the clinical relevance of using NMSK modeling in OA populations. Larger differences were expected between

the predicted KJCFs from each control mode due to their differing method of predicting muscle forces. Although differences occurred within the muscle force predictions, they were not substantial enough to reach significance in the KJCF predictions. Nevertheless, considering both control modes, KJCFs during walking (Fig. 3, Table 2) were similar to those measured by instrumented knee implants following knee replacement surgery for patients with end-stage knee OA (Halder et al., 2012). Discrepancies, especially in medial versus lateral compartment KJCF values (up to 40 % difference), may be attributed to differences in frontal plane alignment and gait speed between the present patients versus those having undergone knee replacement.

This study has limitations. We compared our EMG-driven control mode results to studies that included recorded EMG data from a greater number of muscles. Additionally, we were unable to directly compare our KJCFs to instrumented implant data. Finally, our KJCF results were derived from using a single trial for calibration and four for execution. Including more trials may have strengthened the models' ability to track ID-computed joint moments. Our conclusions, however, are directly comparable to previous studies reporting KJCFs that were verified by instrumented knee implant data within the same population (Hall et al., 2019; S. Starkey et al., 2022; S.C. Starkey et al., 2022).

5. Conclusion

The EMG-assisted control mode had better joint moment tracking accuracy than EMG-driven control mode, and estimated KJCFs comparable to other literature reporting instrumented knee implant data and models of patients with OA. Although the KJCFs did not differ significantly between control modes, our results suggest the EMG-assisted control mode may be more appropriate for evaluating KJCFs in patients with medial-dominant knee OA.

CRedit authorship contribution statement

Dominique Cava: Writing – review & editing, Writing – original draft, Visualization, Project administration, Methodology, Formal analysis, Conceptualization. **Trevor Birmingham:** Writing – review & editing, Supervision, Project administration, Methodology, Funding acquisition, Conceptualization. **Laura E Diamond:** Writing – review & editing, Supervision. **Kristyn Leitch:** Project administration, Investigation. **Ryan Willing:** Writing – review & editing, Supervision, Project administration, Methodology.

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.2025.112773>.

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