



Exposing implausible knee loading: 3D load-balancing reveals limitations in traditional two-step musculoskeletal modeling

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ABSTRACT

Accurate estimation of medial and lateral knee compartment loading is crucial for understanding joint mechanics. The traditional two-step approach, which first estimates musculotendon forces and then derives joint reaction forces and moments, is often followed by a simplified 2D load-balancing method to compute tibiofemoral contact forces in the frontal plane. This study investigated the effect of full 3D knee equilibrium on tibiofemoral contact and ligament force estimation. Gait data from 48 healthy participants were analyzed using a lower-limb musculoskeletal model. The tibiofemoral joint was modeled as a hinge with a moving axis. Joint reaction forces and moments were obtained by subtracting musculotendon forces estimated via static optimization from the intersegmental loads derived through inverse dynamics. Medial and lateral contact forces were then computed either with the 2D or the proposed 3D load-balancing method, which uses all joint reaction load components to estimate ligament and contact forces. The 3D method yielded significantly higher medial and lateral contact forces throughout stance, peaking at 2.50 BW and 1.82 BW, respectively. Ligament forces peaked at 1.05 BW for the ACL, 1.12 BW for the PCL, and 0.44 BW for the LCL, with the MCL largely unloaded. Non-negligible least-squares residuals were observed, indicating limitations in force balance. The 3D load-balancing method revealed physiologically implausible loading patterns arising from joint reaction loads, particularly elevated internal/external rotation moments, computed after independently estimating musculotendon forces in the classical two-step approach. The 2D load-balancing approach is blind to these outcomes. Future work should favor one-step approaches that simultaneously resolve musculotendon, ligament, and contact forces.

1. Introduction

Accurate estimation of medial and lateral knee compartment loading is essential for understanding the mechanical drivers of pathology such as osteoarthritis. Direct in vivo measurement of joint contact forces is restricted to rare cases involving instrumented knee prostheses (Bergmann et al., 2014; D'Lima et al., 2006; Heinlein et al., 2009; Kutzner et al., 2010), and thus applies only to a small number of pathological patients. As a result, non-invasive estimation methods have become a central focus of research and musculoskeletal models of the lower limb are widely used to estimate internal forces during dynamic

tasks. Computational approaches to knee joint loading span a spectrum from perfectly rigid articular surface assumptions to more detailed deformable joint representations (Yang and Pandey, 2024). In deformable, elastic foundation knee models, joint contact mechanics account for the deformation of soft tissues such as cartilage and ligaments (Andersen et al., 2017; Guess, 2012; Guo et al., 2025; Lin et al., 2010; Shelburne et al., 2005). However, such models carry important limitations: the architectural and material properties of joint tissues are difficult to measure in vivo, and their implementation is computationally expensive. Furthermore, yielding appropriate knee contact force estimates does not necessarily guarantee accurate predictions of joint

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kinematics (Guo et al., 2025).

Within rigid-body frameworks, early work proposed simplified formulations (Morrison, 1970, 1968) in which the knee was modelled through a small number of force lines of action, representing muscles, ligaments, and contact forces, together with their associated moment arms. By enforcing moment equilibrium, these models enabled estimation of the internal three-dimensional (3D) load distribution without any penalty-based or deformable modelling. Subsequently, to improve physiological fidelity and address the muscle redundancy problem, researchers turned to optimization techniques or electromyography (EMG)-informed methods (e.g., Delp et al., 1990; Horsman et al., 2007; Moissenet et al., 2014; Rajagopal et al., 2016). In this context, most methods adopt a two-step procedure: musculotendon forces or activations are first estimated, and joint reaction forces and moments are subsequently derived by subtracting these muscle contributions from the intersegmental forces and moments obtained through inverse dynamics (Chèze et al., 2015; Cleather and Bull, 2011, Fig. 1). This approach does not give access to detailed contact and ligament forces. As a consequence, many studies additionally estimate tibiofemoral medial and lateral contact forces using a simplified two-dimensional (2D) load-balancing method (Erdemir et al., 2007; Manal and Buchanan, 2022). This formulation only requires defining the medial and lateral contact point locations, with contact forces assumed to act in the superior-inferior direction. While this computation ensures frontal-plane force equilibrium, it restricts the analysis to this plane, with joint reaction moments beyond the adduction-abduction moment rarely accounted for. However, similarly, previous work has shown that reducing the degrees of freedom in musculotendon force optimization, by idealizing the knee as a planar joint, can generate large unbalanced out-of-sagittal-plane moments (Marouane et al., 2017).

Incorporating additional joint reaction force and moment components within a rigid-body formulation, in the spirit of Morrison's original approach, would enable access to 3D compartmental contact and ligament forces while preserving computational efficiency. The present study aims to investigate the effects of including all components of knee joint reaction forces and moments, computed using a two-step approach, on the estimation of medial and lateral contact forces and ligament forces. These results are compared to those obtained using the conventional 2D frontal-plane formulation, which considers only a subset of the components. We hypothesized that the internal-external rotation reaction moment obtained from the two-step approach cannot be balanced by physiologically realistic tibiofemoral contact and ligament forces. We further hypothesized that this limitation is not apparent in the conventional 2D approach, as it only considers the adduction-abduction moment.

2. Method

2.1. Population

This study is based on a published dataset involving 50 healthy participants (Schreiber and Moissenet, 2019). Participants' gait at overground self-selected speed was recorded using a motion capture system (100 Hz, Oqus4, Qualisys, Sweden) and two force plates (1500 Hz, OR6-5, AMTI, USA). Among the 231 walking trials, 24 were excluded due to incomplete contact of the right foot with the force plate. Two participants (6 trials) were excluded because of missing marker data, and one additional trial was removed due to persistent violations of joint and segment constraints that prevented successful convergence of the kinematic analysis during model setup. A total of 199 stance phases were retained for analysis. The final sample analyzed included 48 participants (26 men and 22 women) with a mean age of 38 ± 14 years, a height of 1.74 ± 0.09 m, and a body mass of 71.6 ± 12.3 kg.

2.2. Musculoskeletal modelling

For each participant, a musculoskeletal model was created based on TLEM 2.0 (Twente Lower Extremity Model v.2.0, Carbone et al., 2015; De Pieri et al., 2018), using the AnyBody Modeling System (AnyBody Technology, Denmark, AMMR, v3.0.4, Lower Extremity mocap model). Body weight and height were used to uniformly scale the model's mass and geometry (Rasmussen et al., 2005). The tibiofemoral joint was modeled as a hinge with a moving axis (Jolas et al., 2025), adapted from Dzialo et al. (2018). In this rigid-body formulation, the time evolution of knee flexion-extension was driven by each participant's gait kinematics obtained from motion capture, whereas the remaining tibiofemoral kinematic degrees of freedom (translations and rotations coupled to flexion) were prescribed from a generic reconstruction of the TLEM 2.0 model. To establish this flexion-dependent kinematic relationship and define the corresponding moving axis, the relative position and orientation of the TLEM 2.0 tibia and femur meshes were reconstructed at 63 discrete tibiofemoral flexion angles (spaced 2° apart) by maximizing joint congruence through contact-pressure minimization (Conconi et al., 2021). Based on this reconstruction, and after interpolating these 63 relative configurations from -5° extension to 120° flexion, the trajectories of the centers of spheres fitted to the contact surfaces between the tibia and the medial and lateral femoral condyles were computed for the entire range of motion. The moving axis was then defined in the model as the line connecting the medial and lateral sphere centers. Consequently, tibiofemoral kinematics were fully prescribed and consisted of a subject-specific flexion-extension motion combined with a generic, flexion-dependent kinematic pattern. No additional subject-specific

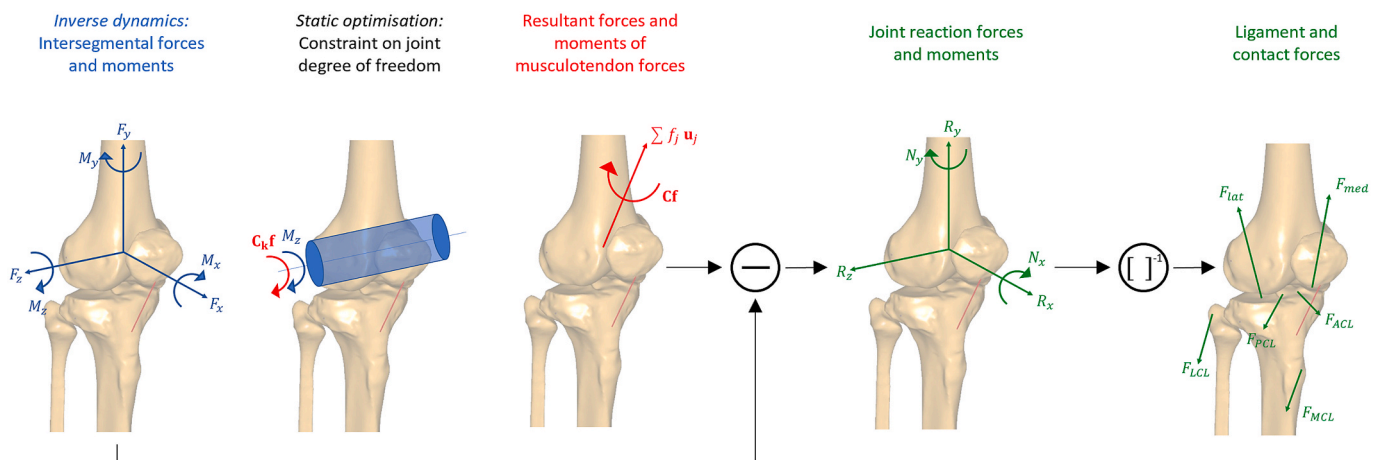


Fig. 1. schematic representation of the two-step approach for the particular case $e_k = Z$

anatomical parameters were included beyond body mass, height, and motion-capture-derived kinematics. Inverse kinematics and inverse dynamics analyses were performed (Andersen et al., 2009). The resulting intersegmental forces and moments, denoted $\begin{bmatrix} \mathbf{F}_i \\ \mathbf{M}_i \end{bmatrix}$ were obtained for each joint i . To resolve the muscle redundancy problem, a static optimization procedure was then applied, in a first step, minimizing a polynomial cost function (Eqs. (1)–(3)) (Damsgaard et al., 2006).

$$\text{Minimize } G(\mathbf{f}) = \sum_{j=1}^n n_j \left(\frac{f_j}{f_{0j}} \right)^3 \quad (1)$$

subject to the dynamic equilibrium constraints

$$\mathbf{C}_k \mathbf{f} = [\dots \mathbf{M}_i \cdot \mathbf{e}_k \dots]^T \quad (2)$$

and the physiological force bounds

$$0 \leq f_j \leq f_{0j}, j \in \{1, \dots, n\} \quad (3)$$

Here, G is the objective function, reflecting the assumed neural control criterion in terms of muscle forces $\mathbf{f}$. The vector $\mathbf{f}$ contains the forces of all musculotendon elements, and n is the total number of muscles in the model. Each f_j denotes the force produced by the j -th muscle, while f_{0j} is its maximum force-generating capacity (muscle strength). The normalization factor n_j , based on the muscle's physiological cross-sectional area, accounts for muscles subdivided into multiple actuators. The ratio $\frac{f_j}{f_{0j}}$ represents the muscle activation level, bounded between zero (no activation) and one (maximum voluntary contraction). The matrix $\mathbf{C}_k$ corresponds to the lever arm matrix which maps the musculotendon forces to intersegmental moments about each rotational degree of freedom, assumed to be controlled by the muscles: three for the hip (spherical joint), one for the knee (hinge joint), and two for the ankle (universal joint). Each k -th rotational degree of freedom is expressed along a unit vector $\mathbf{e}_k$ (e.g., hinge moving-axis for the knee). The right-hand side of the equilibrium equation represents the projection of the intersegmental moments $\mathbf{M}_i$ onto the set of independent degrees of freedom.

Once the optimized muscle forces were obtained, in a second step, the joint reaction forces and moments ($\mathbf{R}_i, \mathbf{N}_i$) were computed as the difference between the intersegmental forces and moments and the musculotendon contribution, as shown in Eq. (4):

$$\begin{bmatrix} \mathbf{R}_i \\ \mathbf{N}_i \end{bmatrix} = \begin{bmatrix} \mathbf{F}_i \\ \mathbf{M}_i \end{bmatrix} - \begin{bmatrix} \sum f_j \mathbf{u}_j \\ \mathbf{C}_i \mathbf{f} \end{bmatrix} \quad (4)$$

where $\mathbf{u}_j$ denotes the unit vector along the line of action of the j -th muscle. The matrix $\mathbf{C}_i$ maps the musculotendon forces to the 3D intersegmental moments. This formulation can be generalized to any joint i and muscles j spanning it. The terms ($\mathbf{R}_i, \mathbf{N}_i$) therefore correspond to the remaining intersegmental joint loads that are not equilibrated by the musculotendon forces and must instead be supported by passive joint structures such as articular contact and ligaments.

2.3. Contact forces estimation

The prescribed joint kinematics and musculotendon forces obtained from the inverse dynamics and static optimization steps were first used to compute the joint reaction forces and moments ($\mathbf{R}_i, \mathbf{N}_i$). At this stage, tibiofemoral contact and ligament forces were not explicitly included in the musculoskeletal model. These computed joint reaction forces and moments were subsequently post-processed a posteriori (MATLAB version R2024b, MathWorks, Natick, MA, USA) to determine the tibiofemoral medial and lateral contact and ligament forces through a 3D static equilibrium analysis. Knee contact point trajectories, surface normals, and ligament attachment sites on the femur and tibia were

obtained from the aforementioned congruence computation applied to the TLEM 2.0 tibia and femur meshes (Conconi et al., 2021; Dumas et al., 2024). It was checked that the distances between femoral and tibial attachments remain almost constant during the flexion simulated with the congruence computation. Medial and lateral contact points were defined as the centroids of their respective contact areas. Contact areas were identified as regions of the articular surfaces with inter-surface distances below 2 mm. Contact normals were calculated as weighted averages of local surface normals, with weights based on surface proximity (Conconi et al., 2014). The direction of each contact force was assumed to align with its corresponding surface normal. Ligament forces for the anterior cruciate (ACL), posterior cruciate (PCL), medial collateral (MCL), and lateral collateral (LCL) ligaments were applied at their tibial insertions, directed along the line connecting the femoral and tibial attachments, consistent with a tensile force acting on the tibia. Ligament origins, insertions, lines of action, and moment arms were determined from the prescribed tibiofemoral kinematics at each time step. This simplified representation was adopted to focus the analysis on the mechanical consistency of the 3D load-balancing method rather than detailed subject-specific ligament modelling.

Subsequently, medial and lateral contact forces, and, when relevant, ligament forces, were estimated using two methods (illustrated in Fig. 2):

- **3D:** the amplitudes of the medial and lateral contact forces and those of the ACL, PCL, MCL, and LCL (g_i), were estimated via a least-squares optimization (Eq. 5). This optimization integrated their corresponding lines of action ($\mathbf{u}_i$) and lever arms ($\mathbf{L}_i$) and the previously computed joint reaction forces and moments ($\mathbf{R}_i, \mathbf{N}_i$).

$$\min \left\| \begin{bmatrix} \mathbf{u}_1 \dots \mathbf{u}_6 \\ \mathbf{L}_1 \dots \mathbf{L}_6 \end{bmatrix} \begin{bmatrix} g_1 \\ \vdots \\ g_6 \end{bmatrix} - \begin{bmatrix} \mathbf{R}_i \\ \mathbf{N}_i \end{bmatrix} \right\|^2 \quad (5)$$

subject to $g_i > 0$ to ensure tensile-only behavior of ligaments and compressive-only behavior of contacts.

- **2D:** medial and lateral contact forces along the superior-inferior direction (F_{med-y}, F_{lat-y}) were estimated based on their medial-lateral lever arms (L_{med-x}, L_{lat-x}), the superior-inferior component of the joint reaction force (R_y), and the abduction-adduction joint reaction moment (N_x) (Eq. (6)).

$$\min \left\| \begin{bmatrix} 1 & 1 \\ L_{med-x} & -L_{lat-x} \end{bmatrix} \begin{bmatrix} F_{med-y} \\ F_{lat-y} \end{bmatrix} - \begin{bmatrix} R_y \\ N_x \end{bmatrix} \right\|^2 \quad (6)$$

subject to $(F_{med-y}, F_{lat-y}) > 0$ to ensure compressive-only behavior of contacts

2.4. Data processing and statistical analysis

Residuals were computed for both the 3D and 2D least-squares methods. Medial and lateral contact and ligament forces were normalized to body weight (BW), and joint reaction moments were normalized to body weight multiplied by leg length (BW-leg length). Differences between the 3D and 2D load-balancing methods in estimating medial and lateral contact forces were assessed across the entire stance phase using statistical parametric mapping (SPM) with a paired t -test, implemented via the `spm1d` package (version M.0.4.8, in MATLAB; Pataky, 2012). Statistical significance was set at $p < 0.05$. Significant differences between methods reflect phases of the stance cycle where considering the full 3D joint reaction loads alters the redistribution of contact forces compared with the simplified 2D formulation. These differences should be interpreted as consequences of the modelling assumptions underlying each approach rather than as evidence that one method is more valid than the other.

All estimated forces and moments were expressed in a tibial

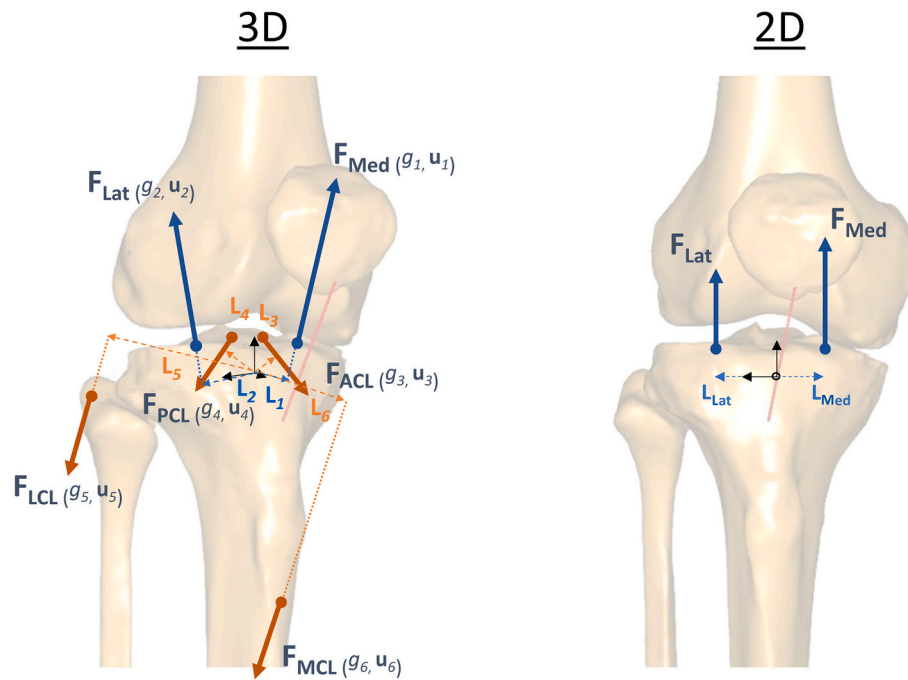


Fig. 2. Overview of the two computational methods (corresponding to Eqs. 5 and 6) used to estimate tibiofemoral contact and ligament forces. **3D** - F_{Med}/F_{Lat} denote the medial and lateral contact forces; F_{PCL}/F_{ACL} the posterior and anterior cruciate ligament forces; and F_{LCL}/F_{MCL} the lateral and medial collateral ligament forces. g_i indicates the force amplitudes, u_i the force directions, and L_i the lever arms, with $i = 1:6$ corresponding to these six force-generating structures. **2D** - F_{Med} and F_{Lat} denote the medial and lateral contact forces, and L_{Lat}/L_{Med} their medio-lateral respective lever arms

coordinate system: the origin was placed at the midpoint between the most medial point of the medial tibial condyle and the most lateral point of the lateral tibial condyle. The Z-axis was defined along the line connecting these two points (pointing laterally). The X-axis was defined as the direction orthogonal to the plane formed by the Z-axis and the long axis of the shank (pointing anteriorly), and the Y-axis as the cross-product $Y = Z \times X$ (pointing superiorly). Axes X, Y, and Z are interpreted as antero-posterior/adduction-abduction, superior-inferior/

internal-external rotation, and medial-lateral/flexion-extension, respectively. Note that the tibial-frame flexion-extension used for reporting moments does not coincide with the joint's actual flexion-extension, which in the model occurs about a moving hinge axis corresponding to the knee's functional rotation. This distinction only affects the representation of the reported quantities and does not influence the underlying joint mechanics or force calculations.

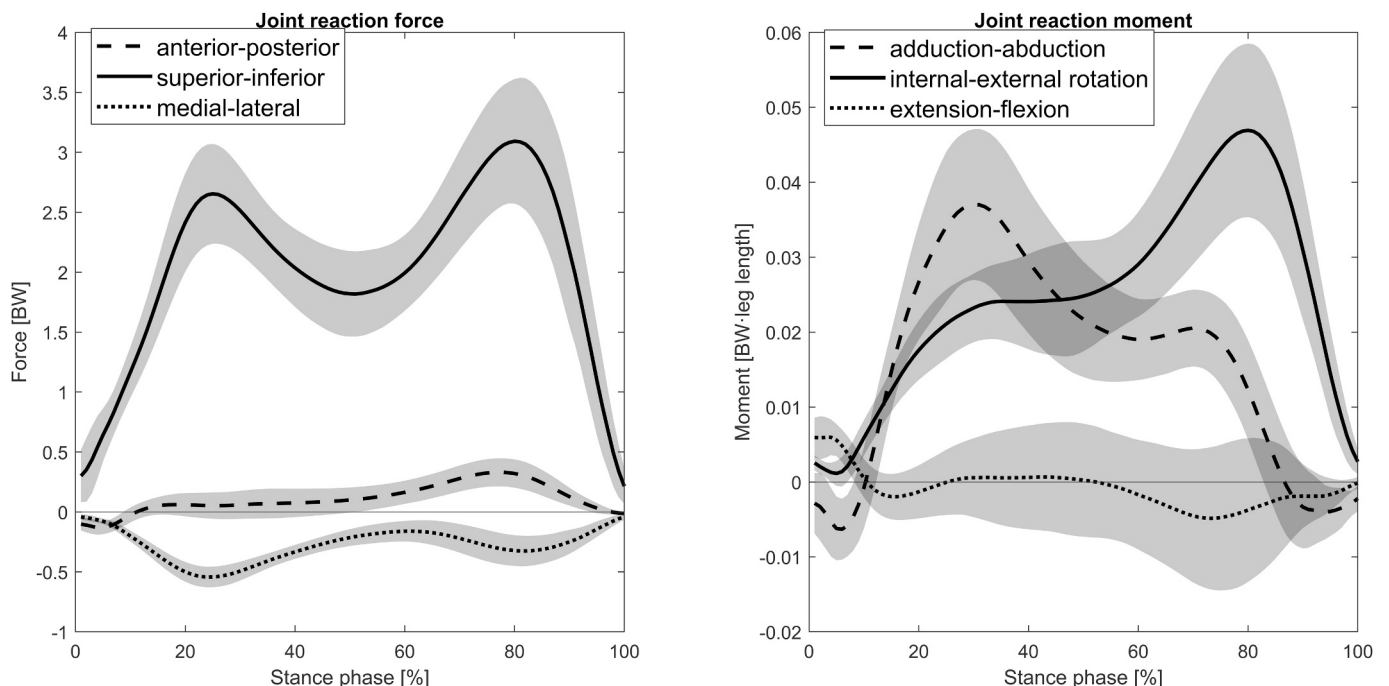


Fig. 3. Joint reaction forces and moments computed following static optimization. Shaded areas: the standard deviations for each force/moment

3. Results

3.1. Joint reaction forces and moments

The average joint reaction forces and moments obtained from the musculoskeletal model, used as inputs to estimate contact (and ligament) forces, are expressed in the tibial coordinate system (Fig. 3). The superior-inferior joint reaction force reached a peak of 3.09 ± 0.52 BW at 80 % of the stance phase (Table 1), while the peak adduction-abduction and internal-external rotation moments reached 0.037 ± 0.010 BW-leg length at 31 % of the stance phase and 0.047 ± 0.012 BW-leg length at 80 % of the stance phase, respectively.

Comparison of 3D and 2D contact force estimations

Fig. 4 presents a comparison of medial and lateral contact force magnitudes estimated using the 3D and 2D load-balancing methods. Fig. 5 shows the corresponding residual joint reaction forces and moments obtained from the 3D load-balancing approach, reflecting the portion of the joint reaction quantities that could not be balanced within the imposed modelling constraints. Table 2 presents peak knee contact and ligament forces estimated using 2D and 3D load-balancing methods, compared with similar studies. In 27 out of 199 trials, abnormal force peaks (up to 10.7 BW) were observed at 96 ± 3 % of the stance phase when using the 3D method. The 3D method produced, on average, significantly higher contact forces from 0 to 99 % of the stance phase for the medial compartment, and from 0 to 10 %, 21–90 %, and 92–99 % of the stance phase for the lateral compartment, compared to the 2D method, reaching up to 2.50 ± 0.49 BW at 78 % of the stance phase and 1.82 ± 0.41 BW at 82 % of the stance phase, respectively. Estimated ligament forces peaked at 1.05 ± 0.35 BW for the ACL, 1.12 ± 0.32 BW for the PCL, and 0.45 ± 0.18 BW for the LCL, while the MCL remained unloaded during most of the stance phase. On average, joint reaction moment residuals peaked at 25.2 % at the beginning of the stance phase for the adduction-abduction component, then remained close to zero. With the 3D load-balancing method, residuals reached 0.006 ± 0.003 BW-leg length (15 % of the peak-to-peak amplitude) for the adduction-abduction component, -0.0219 ± 0.010 BW-leg length (−48 %) for the internal-external rotation component, and 0.0145 ± 0.007 BW-leg length (128 %) for the flexion–extension component. Residuals for joint reaction forces remained below ± 0.002 BW (± 0.4 %) throughout the stance phase. The residuals were consistently negligible in the 2D force-balancing method.

4. Discussion

This study aimed to propose a 3D load-balancing method for estimating tibiofemoral medial and lateral contact and ligament forces from the joint reaction forces classically obtained through the two-step

approach, and to compare it with a commonly used 2D method. Directions of the contact normals and ligament lines of action as well as lever arms had to be defined for the 3D method. The abnormal force peaks consistently occurring at a given instant of the stance phase seem to demonstrate that this specific geometry, based on the TLEM 2.0 and consistent with the moving axis definition, can eventually reach a singular (or close to singular) configuration.

Our results show that the 3D load-balancing method can result in non-physiological estimates of medial and lateral contact and ligament forces and non-negligible least-squares residuals. In this case, the issue does not stem from a singular configuration but reveals that all the components of joint reaction loads cannot be balanced by a set of forces matching the tibiofemoral anatomy. This was reflected in excessively high ligament and contact forces, with ACL forces at least two times higher than values typically reported in more anatomically detailed knee models (Charles et al., 2021; Frigo and Donno, 2021; Hu et al., 2018; Kia et al., 2014; Shao et al., 2011; Shelburne et al., 2005; Vignos et al., 2020, Table 2) and approaching experimentally reported failure thresholds (Woo et al., 1991). They also revealed markedly increased residual moments around the flexion–extension and internal-external rotation axes in the fixed tibial reference frame. These residuals reflect the unbalanced portion of the joint reaction moments that cannot be redistributed to contact and ligament forces under the modelling assumptions (i.e., imposed force directions, application points, and positivity constraints). Such inconsistencies may remain undetected when relying on a 2D load-balancing approach.

These excessively high forces likely arise from non-physiological joint reaction loads. In particular, the internal-external rotation moment predicted by our model is at least three times higher than values reported in studies using instrumented knee implants (Fregly et al., 2012; Kutzner et al., 2010; Taylor et al., 2017), and even exceeds the abduction–adduction moment, contrary to in vivo observations (Table 1). Although the 2D load-balancing method may yield seemingly reasonable results due to its simplification, it neglects other joint reaction components, beyond the superior–inferior force and adduction–abduction moment, that, as observed in our tibiofemoral model, are more challenging for the knee passive structures to equilibrate. These limitations stem from the classical two-step approach, in which musculotendon forces are estimated independently of joint contact and ligament interactions. Modeling the tibiofemoral joint as a hinge implies that it is infinitely stiff against all loads and moments except for the extension–flexion. Even in the present study, where the axis of extension–flexion is a moving axis built in a consistent way with the contact and ligament geometry, it is not ensured that the balanced (with the extension–flexion moment) and minimized musculotendon forces also yield balanced and minimized contact and ligament forces. Although inter-individual and sex-related differences may influence knee mechanics, the present study focuses on the mechanical consistency of joint reaction loads generated by the classical two-step approach. In this

Table 1
Peak knee joint reaction forces and moments: comparison with instrumented implant studies.

Study	Joint reaction force			Joint reaction moment		
	Anterior-posterior	Superior-inferior	Medial-lateral	Adduction-abduction	Internal-external rotation	Extension-flexion
Kutzner et al., 2010	0.28 BW	2.6 BW	0.12 BW	0.030 Nm/BW	0.011 Nm/BW	0.019 Nm/BW
Fregly et al., 2012 (Grand Challenge) values from Kia et al., 2014	100 N	2050 N	100 N	10 Nm	2 Nm	20 Nm
Taylor et al., 2017 (CAMS-Knee)	0.23 BW	2.5 BW	0.07 BW	0.04 Nm/BW	0.007 Nm/BW	0.021 Nm/BW
Present study	0.33 ± 0.12 BW 235 ± 92 N	3.09 ± 0.52 BW 2182 ± 444 N	0.54 ± 0.09 BW 385 ± 88 N	0.0303 ± 0.008 Nm/ BW 21.3 ± 5.8 Nm	0.0383 ± 0.008 Nm/BW 27.0 ± 6.6 Nm	0.005 ± 0.002 Nm/ BW 3.53 ± 1.61 Nm

BW: body weight.

Some values were approximated from digitized curves. All values are reported in their original units (e.g., BW or N), while joint moments from our study were normalized to Nm/BW to enable comparison.

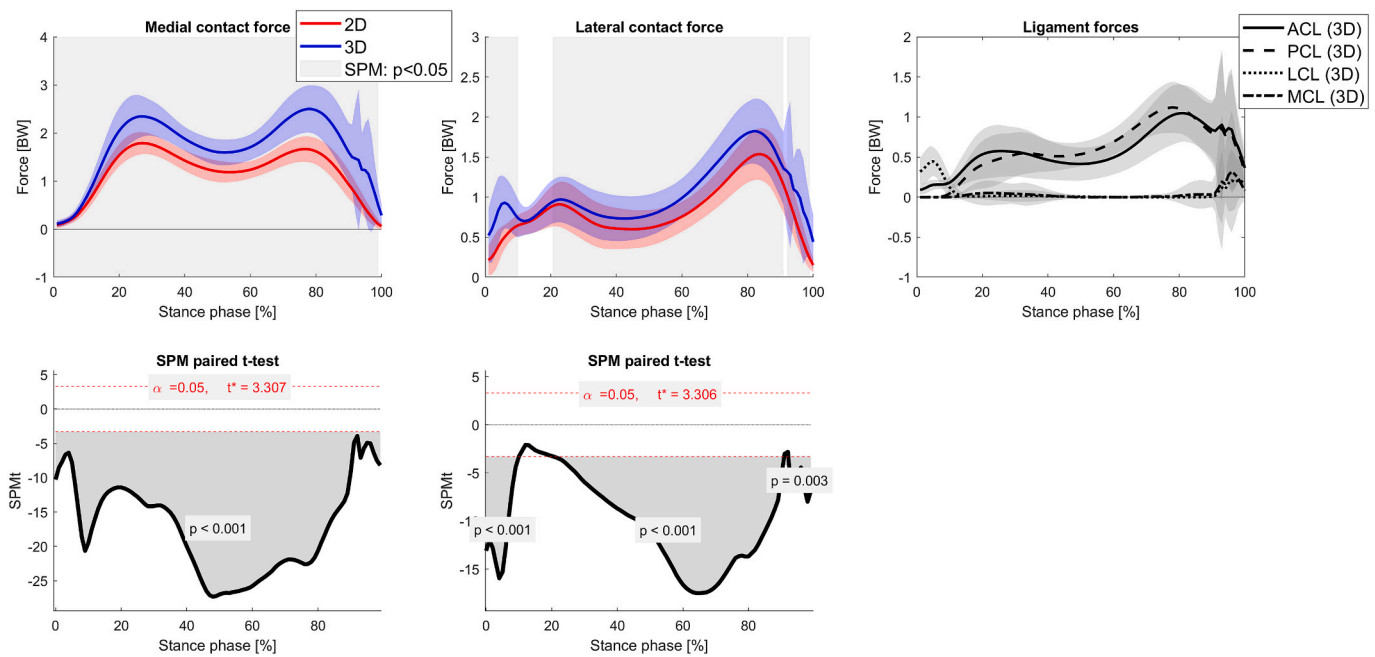


Fig. 4. Comparison of medial and lateral contact forces computed using 3D (blue continuous line) and 2D (red continuous line) load-balancing method. Shaded areas: the standard deviations for each force/moment. SPM: statistical parametric mapping. ACL: anterior cruciate ligament, PCL: posterior cruciate ligament, MCL: medial collateral ligament, LCL: lateral collateral ligament

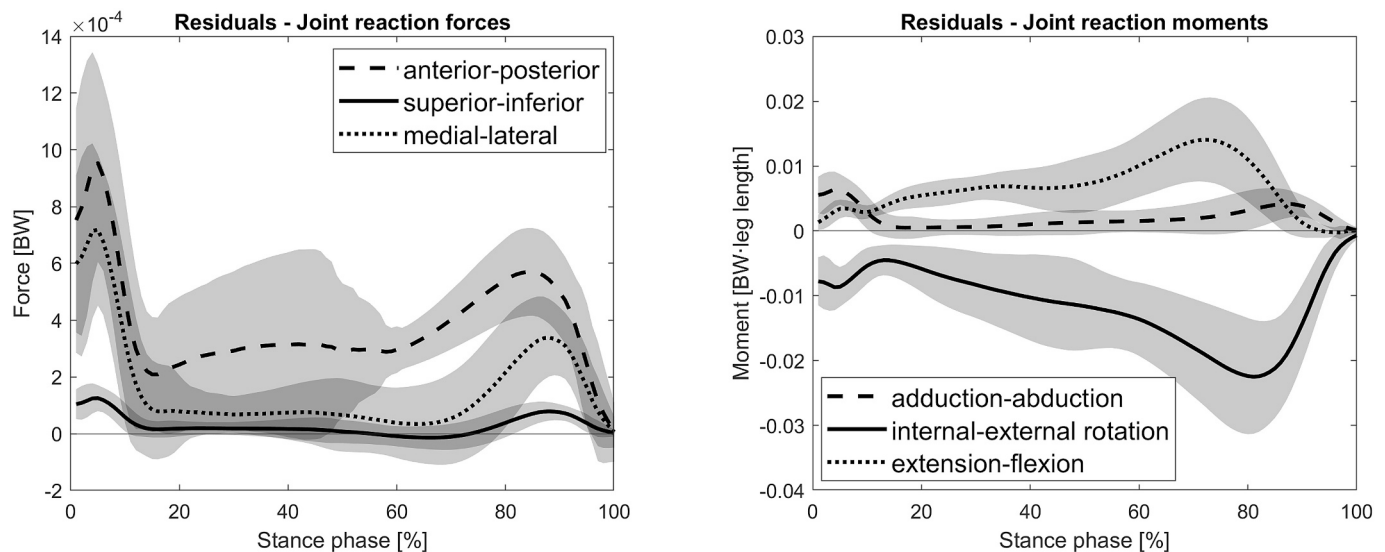


Fig. 5. Joint reaction force and moment least-squares residuals (unbalanced in 3D equilibrium). BW: body weight

context, given the magnitude and nature of these loads, particularly the internal-external rotation moment, the observed equilibrium inconsistencies are primarily driven by the modelling framework itself and are expected to persist across individuals.

A phenomenon illustrating this lack of physiological consistency in computed joint reaction forces and moments is the so-called *lateral unloading*, observed in the results of some 2D load-balancing methods (Kumar et al., 2013; Shelburne et al., 2006), where lateral contact forces become negative during stance. This occurs primarily due to the external knee adduction moment, which provides a negative contribution to the lateral compartment load (Zhang et al., 2025). In reality, muscle (and ligament) forces act to counterbalance this effect by stabilizing the joint. Although muscles such as the quadriceps, hamstrings, gastrocnemius, and tensor fasciae latae are mainly involved in sagittal-plane propulsion,

they also contribute significantly to frontal-plane stabilization (Kumar et al., 2013; Shelburne et al., 2006; Winby et al., 2009). However, static optimization frameworks that minimize muscle activations without incorporating additional stabilization constraints often fail to reproduce these mechanisms. A parallel can be made with upper limb musculo-skeletal models, including the glenohumeral joint. The contact force is often constrained to point into the glenoid cavity (Charlton and Johnson, 2006). Removing such constraints highly alters musculo-tendon force predictions and can even lead to a joint reaction force oriented toward dislocation. Similarly, introducing additional constraints into tibiofemoral models could help prevent non-physiological outcomes in medial and lateral contact mechanics as well as in ligament force estimations. Alternatively, strain-driven approaches prescribe ligament forces directly from joint kinematics (Kotsifaki et al.,

Table 2

Peak knee contact and ligament forces: comparison with other musculoskeletal modelling studies.

Study		Total force	Medial Force	Lateral Force	ACL	PCL	MCL	LCL
Shelburne et al., 2005		2050 N	1700 N	500 N	300 N	100 N	20 N	125 N
Shao et al. 2011		—	—	—	0.37 BW	—	0.19 BW	0.39 BW
Hu et al. 2018		4.7B W	2.8 BW	1.9 BW	30 N	195 N	150 N	155 N
Frigo & Donno, 2021		2300 N	—	—	205 N	680 N	100 N	120 N
Charles et al., 2021		3.0 BW	—	—	0.43 BW	—	—	—
Vignos et al., 2020		—	—	—	170 N	—	—	—
Kia et al., 2014		—	500 N	1100 N	—	80 N	180 N	200 N
Present study	2D	3.09 ± 0.52 BW	1.79 ± 0.23 BW	1.53 ± 0.33 BW	—	—	—	—
		2183 ± 444 N	1264 ± 220 N	1082 ± 248 N	—	—	—	—
	3D	4.27 ± 0.87 BW	2.50 ± 0.49 BW	1.82 ± 0.41 BW	1.05 ± 0.35 BW	1.12 ± 0.32 BW	0.45 ± 0.18 BW	0.32 ± 0.76 BW
		3001 ± 623 N	1757 ± 361 N	1280 ± 244 N	736 ± 249 N	785 ± 219 N	314 ± 126 N	219 ± 533 N

Some reported values are approximated from digitized curves. All values are reported in their original units (e.g., BW or N).

BW: body weight, ACL: anterior cruciate ligament, PCL: posterior cruciate ligament, MCL: medial collateral ligament, LCL: lateral collateral ligament.

2022; Smale et al., 2019). While this may avoid explicit equilibrium inconsistencies, these methods remain highly sensitive to kinematic constraints (e.g., joint degrees of freedom, coupling definitions, soft-tissue artefact compensation...), as well as ligament parameters such as slack length and stiffness. In addition, they do not explicitly consider mechanical interaction between muscle, ligament, and contact forces.

A promising direction involves solving for musculotendon, ligament, and contact forces concurrently within the one-step approach (Cleather and Bull, 2011; Eskinazi and Fregly, 2018; Guess, 2012; Manal and Buchanan, 2022; Moissenet et al., 2014). EMG-driven models also provide an alternative to purely optimization-based approaches (Ao et al., 2022; Crossley et al., 2025; Princelle et al., 2025; Saxby et al., 2016). Some studies even combine both approaches by integrating EMG-driven inputs within one-step formulations that include additional constraints, for instance, modulating the medial-to-lateral force distribution through a penalty function (Manal and Buchanan, 2022). By explicitly minimizing contact and ligament forces, such methods could promote more physiologically realistic muscle recruitment strategies, enhancing the local stabilizing role of muscles and their interplay with passive structures while maintaining global dynamic equilibrium.

This study has limitations. First, accurately defining ligament insertion sites is challenging, as these structures often attach over broad regions rather than at discrete points. Moreover, their lines of action were simplified to single straight vectors between origin and insertion points, whereas real ligament anatomy includes multiple functional bundles with varying orientations (Logterman et al., 2018; Petersen and Zantop, 2007). Second, only the primary knee ligaments were included in the analysis. Other passive stabilizing structures such as the joint capsule, menisci, and smaller extra-capsular ligaments were not accounted for. These oversimplifications may misrepresent force magnitudes, points of application, and directions. Third, the model was based on a generic musculoskeletal template that ensures internal consistency across anatomical structures (e.g., contact geometry, ligament and muscle paths, joint axes...). However, the external loading applied in simulations was subject-specific, which may introduce mismatches between model structure and biomechanical input. Finally, the large force peaks observed in the 3D load-balancing method with inclusion of ligaments likely reflect geometric near-singular configurations, occurring when the determinant of the passive-structure force-direction matrix approaches zero. Such near-singularities arise at higher knee flexion, particularly during the final 10 % of the stance phase, when the lines of action of the ligaments and contact forces become nearly collinear. In these conditions, small perturbations in joint reactions or geometry produce disproportionately large variations in the estimated passive forces, resulting in artificially amplified passive force predictions. Such singular configurations are well-known and generally taken into account in the synthesis of knee models with punctual contacts and a limited number of isometric ligaments (Barzan et al., 2019). These configurations highlight the need for caution when interpreting passive force estimates in deeper flexion and warrant further

investigation into more detailed tibiofemoral models and singularity-robust formulations. While these limitations are acknowledged, the aim of this study is primarily to highlight a methodological issue, as the observed results are mainly driven by the joint reaction force and moment estimation framework within the classical two-step approach, rather than by the level of anatomical detail or subject-specific fidelity in the representation of ligament and other musculoskeletal structures.

In conclusion, this work highlights that extending load-balancing approaches to a full 3D formulation after the classical two-step approaches, i.e., based solely on musculotendon forces minimization, reveals inconsistencies in the estimated joint reaction forces and moments. In particular, the resulting contact and ligament force distributions demonstrated that joint reaction loads computed without considering the complex interplay among muscles, ligaments, and joint contact mechanics may yield physiologically implausible loading patterns, potentially leading to misleading interpretations. To address these limitations, future research should prioritize one-step modeling approaches that simultaneously resolve musculotendon, ligament, and contact forces. Moreover, improving the physiological consistency of joint loading requires explicitly defining ligament paths to ensure their coherent mechanical contribution to knee stabilization. Such approaches are better positioned to capture the complexity of articular mechanics and to provide more reliable insights into joint loading, particularly in clinical contexts such as joint degeneration, gait abnormalities, or presurgical planning.

CRediT authorship contribution statement

Elisa Jolas: Writing – original draft, Visualization, Software, Methodology, Formal analysis, Conceptualization. **Sacha Guitteny:** Writing – review & editing, Supervision, Methodology, Formal analysis, Conceptualization. **Nolwenn Fougeron:** Writing – review & editing, Supervision, Conceptualization. **Rachid Aissaoui:** Writing – review & editing, Supervision, Conceptualization. **Charles Pontonnier:** Writing – review & editing, Supervision, Conceptualization. **Raphaël Dumas:** Writing – review & editing, Supervision, Methodology, Formal analysis, Conceptualization.

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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