

Article

Inverse Dynamics Analysis of Knee Joint Kinetics During Squatting: A Cross-Sectional Comparative Study Between Native Healthy Knees and Posterior-Stabilized Total Knee Arthroplasty

Kasiwat Chaiyasit ¹, Chotchuang Phombut ² and Supakit Rooppakhun ^{2,*} 

¹ School of Biomedical Innovation Engineering, Institute of Engineering, Suranaree University of Technology, Nakhon Ratchasima 30000, Thailand; kasiwat.chai@gmail.com

² School of Mechanical Engineering, Institute of Engineering, Suranaree University of Technology, Nakhon Ratchasima 30000, Thailand; chotchuang.p@gmail.com

* Correspondence: supakit@sut.ac.th

Abstract

A biomechanical evaluation of knee loading during squatting is essential for understanding functional capacity after total knee arthroplasty (TKA). This study compares knee joint kinetics in healthy native knees and in posterior-stabilized TKA (PS-TKA) across BMI categories using 3D motion capture and inverse dynamics. Sixty-two knees (31 healthy, 31 PS-TKA) were analyzed. Native knees demonstrated greater flexion capacity and higher joint loading than PS-TKA knees. Peak resultant joint forces reached 3.50 ± 1.00 BW in healthy knees compared with 2.90 ± 1.20 BW in PS-TKA knees. Healthy knees also generated higher joint moments, with maximum adduction and rotation moments of $5.07\% \text{ BW} \times \text{height}$ and $1.29\% \text{ BW} \times \text{height}$, respectively. Body mass index (BMI) significantly influenced loading patterns in native knees, increasing anterior–posterior forces, quadriceps demand, and resultant moments, whereas loading in PS-TKA knees showed minimal BMI dependence. These findings highlight fundamental biomechanical differences between native and prosthetic knees and provide population-specific insights relevant to rehabilitation and high-flexion activities common in Asian populations.

Keywords: knee kinetics; squatting biomechanics; inverse dynamics; motion analysis; total knee arthroplasty; musculoskeletal modeling

1. Introduction

The knee is one of the most complex load-bearing joints in the human body, contributing to balance and mobility during daily functional activities [1]. Furthermore, recent biomechanical evidence suggests that optimizing movement strategies, particularly during landing and high-impact tasks, is essential for reducing the risk of lower-limb injury [2]. Typical activities of daily living require knee flexion of approximately $20\text{--}120^\circ$ [3–5]. In Asian populations, however, squatting is common and often involves knee flexion exceeding 130° , with reported angles reaching 160° during tasks such as toileting, floor sitting, and meditation [6–9]. Recent kinematic studies have highlighted that squatting styles and depths differ significantly between Asian and Western populations. Yet most total knee arthroplasty (TKA) designs and rehabilitation protocols are based primarily on Western biomechanical standards [10].

Deep or repetitive squatting imposes substantial tibiofemoral loading and has been associated with increased mechanical stress, altered cartilage sliding, and elevated risk of



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degenerative changes [11–14]. For patients undergoing TKA, achieving such high flexion safely is a major challenge. Retrospective analyses indicate that TKA patients often adopt altered squatting positions or fail to attain deep flexion due to prosthetic design limitations or soft-tissue tightness [15]. Understanding the kinetic demands of these compensatory movements is crucial for preventing implant wear and loosening.

Body mass index (BMI) is another key factor influencing knee biomechanics. Elevated BMI increases tibiofemoral contact forces, modifies gait mechanics, and accelerates joint degeneration in both healthy and osteoarthritic populations [16–21]. Excess body mass further alters postural strategies and muscular demand during weight-bearing movements, potentially affecting knee loading during high-flexion tasks such as squatting. However, the interplay between high BMI and the mechanical constraints of TKA implants during deep squatting remains insufficiently explored. While obesity is a known predictor of adverse TKA outcomes [22], few studies have quantified how excess body mass specifically alters joint kinetics during deep flexion and high-load conditions.

From an engineering perspective, accurate quantification of joint loads is essential for optimizing implant geometry and predicting material failure. Three-dimensional motion capture provides a precise, noninvasive method for quantifying human movement with sub-millimeter accuracy [23,24]. When combined with force plates, inverse dynamics enables the computation of joint reaction forces, moments, and muscle contributions [25–28]. Furthermore, the integration of these datasets with advanced musculoskeletal modeling has seen significant progress in recent years. Validated subject-specific models now enable reliable prediction of in vivo contact forces without invasive instrumentation [29,30]. These computational tools are becoming indispensable for “digital twin” applications in orthopedics, enabling the assessment of implant performance under realistic, patient-specific loading conditions.

Posterior-stabilized total knee arthroplasty (PS-TKA) is designed to replicate posterior cruciate ligament function and provide stable kinematics throughout flexion. Nevertheless, multiple studies have reported that PS-TKA knees exhibit altered kinematics and reduced deep flexion compared with healthy native knees [31–33]. Differences in muscular recruitment, ligamentous constraints, and prosthesis geometry may influence force transmission during squatting, making this movement particularly relevant for postoperative evaluation. Currently, there is a lack of comparative data integrating kinetics, high-flexion biomechanics, and BMI analysis within a single study framework. Addressing this gap is vital for developing next-generation implants and AI-driven rehabilitation monitoring systems that can accommodate the specific needs of high-BMI and high-flexion populations [34].

Following PICO principles, this study aimed to compare knee joint kinetics (Outcome) between native healthy knees and PS-TKA knees (Population/Comparison) during deep squatting (Intervention), with a specific focus on the modifying effect of Body Mass Index (BMI). We hypothesized that while native knees would demonstrate load modulation proportional to BMI, PS-TKA knees would exhibit altered loading patterns due to prosthetic mechanical constraints.

2. Materials and Methods

2.1. Participants and Data Acquisition

This study involved two distinct participant groups: healthy young adults with native knees and elderly patients who had undergone posterior-stabilized total knee arthroplasty (PS-TKA). The study included 62 knees; the healthy group consisted of 31 native knees (21 males, 10 females; mean age 24.8 ± 2.2 years; BMI 25.14 ± 6.65 kg/m²) with no history of knee injury, surgery, or musculoskeletal disorders. The PS-TKA group comprised 31 knees (7 males, 24 females; mean age 66.5 ± 6.8 years; BMI 28.05 ± 4.28 kg/m²). The

same orthopedic surgeon performed all PS-TKA procedures, and clinical assessments were conducted at least one year post-operatively to ensure complete functional recovery. Participants were excluded if they had structural deformities, neuromuscular disorders, active infections, or acute knee injuries at the time of testing.

The study protocol was approved by the Human Research Ethics Committee (Approval No. 97/2561), and written informed consent was obtained from all participants before data collection. To analyze the effect of body mass, participants were stratified into three BMI categories (light, middle, and heavy) based on the Asia-Pacific BMI classification [35]. Grouping thresholds were verified using K-means clustering analysis to ensure optimal separation [36]. Consequently, a BMI of $\geq 23 \text{ kg/m}^2$ was considered overweight, and $\geq 25 \text{ kg/m}^2$ was classified as obese in this specific population context. The BMI distributions for each group are detailed in Table 1.

Table 1. BMI classification of study participants.

Parameter	Normal ($n = 31$) Mean $\pm$ SD	TKA ($n = 31$) Mean $\pm$ SD
Light	19.40 $\pm$ 2.25 ($n = 11$)	21.80 $\pm$ 7.66 ($n = 11$)
Middle	26.01 $\pm$ 3.51 ($n = 16$)	27.73 $\pm$ 2.53 ($n = 15$)
Heavy	37.44 $\pm$ 6.14 ($n = 5$)	37.73 $\pm$ 2.53 ($n = 5$)

Before motion capture, 41 retro-reflective spherical markers were placed on specific anatomical landmarks according to the Plug-in-Gait marker placement protocol [37], ensuring accurate tracking of the pelvis, thigh, shank, and foot segments (Figure 1).



Figure 1. Placement of 41 reflective markers at anatomical landmarks according to the Plug-in-Gait protocol for capturing pelvis, thigh, shank, and foot kinematics. Abbreviations: R/L = Right/Left; FHD = Front Head; BHD = Back Head; C7 = 7th Cervical Vertebra; T10 = 10th Thoracic Vertebra; CLAV = Clavicle; STRN = Sternum; BAK = Back; SHO = Shoulder; UPA = Upper Arm; ELB = Elbow; FRA = Forearm; WRA/WRB = Wrist; FIN = Finger; ASI = Anterior Superior Iliac Spine; PSI = Posterior Superior Iliac Spine; THI = Thigh; KNE = Knee; TIB = Tibia; ANK = Ankle; HEE = Heel; TOE = Toe; MT5 = 5th Metatarsal.

2.2. Motion Capture and Musculoskeletal Modeling System

Kinematic data were acquired using a six-camera infrared motion-capture system (Oqus 700, Qualisys Inc., Gothenburg, Sweden) at a sampling frequency of 150 Hz. Synchronized ground reaction force (GRF) data were collected using force plates (Type 5233A, Kistler Instrument AG, Winterthur, Switzerland; dimensions 40 cm $\times$ 60 cm) embedded

in a flat floor and sampled at 1000 Hz. Marker trajectories were filtered using a 4th-order Butterworth low-pass filter at 6 Hz, while GRF data were filtered at 20 Hz. Motion data were processed using Qualisys Track Manager (QTM) software version 2.15 (Qualisys AB, Gothenburg, Sweden) and subsequently exported to the AnyBody Modeling System (AMS) version 7.3.1.8217 (AnyBody Technology, Aalborg, Denmark).

A personalized full-body musculoskeletal model was constructed using the GaitFull-Body model from the AnyBody Managed Model Repository (AMMR v2.2.3). This system integrated 3D motion-capture data, force-plate measurements, and subject-specific anthropometric scaling (based on age, gender, body mass, and height) to compute joint reaction forces, moments, and muscle forces via inverse dynamics. Figure 2 illustrates the complete workflow from data acquisition to kinetic computation.

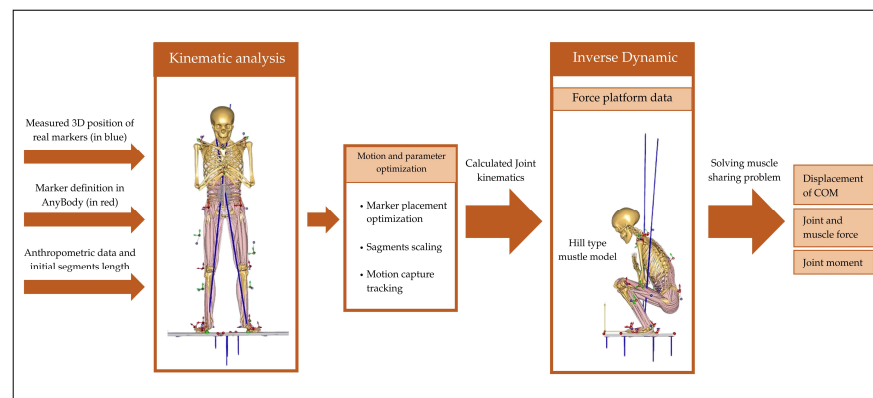


Figure 2. Workflow of 3D motion capture and inverse dynamics analysis. Orange arrows indicate the direction of the workflow. In the kinematic analysis, blue spheres represent measured 3D positions of real markers, while red spheres represent marker definitions in AnyBody. In the inverse dynamic model, blue lines represent ground reaction force vectors.

2.3. Inverse Dynamics and Muscle Recruitment Strategy

The musculoskeletal analysis focused on lower-limb kinetics, using a 3D coordinate system to describe joint dynamics based on anatomical landmarks and axes (X, Y, Z) (Figure 3). To address the muscle redundancy problem—where the number of muscles exceeds the number of equilibrium equations—the study employed an inverse-dynamics optimization approach. The problem was formulated to minimize a specific objective function, G , subject to mechanical constraints:

$$\text{Minimize } G(f^{(M)}) \quad \text{Subject to : } \mathbf{C}\mathbf{f} = \mathbf{d} \\ 0 \leq f_i^{(M)} \leq N_i, \quad i \in \{1, \dots, n^{(M)}\}$$

where

- $f_i^{(M)}$ represents the force of the i^{th} muscle.
- N_i is the physiological strength limit of the muscle, derived from the Hill-type muscle model parameters within the AMMR dataset.
- $\mathbf{C}$ is the coefficient matrix of the equilibrium equations.
- $\mathbf{d}$ is the vector of external loads, including inertial and intersegmental forces.

In this study, a polynomial muscle recruitment criterion was adopted as the objective function to simulate the central nervous system's strategy for minimizing muscle fatigue:

$$G(f^{(M)}) = \sum_{i=1}^{n^{(M)}} \left(\frac{f_i^{(M)}}{N_i} \right)^2$$

This criterion minimizes the sum of cubed muscle activities, yielding a physiologically relevant estimate of muscle forces (e.g., quadriceps force) and joint contact loads, consistent with validated approaches [38,39].

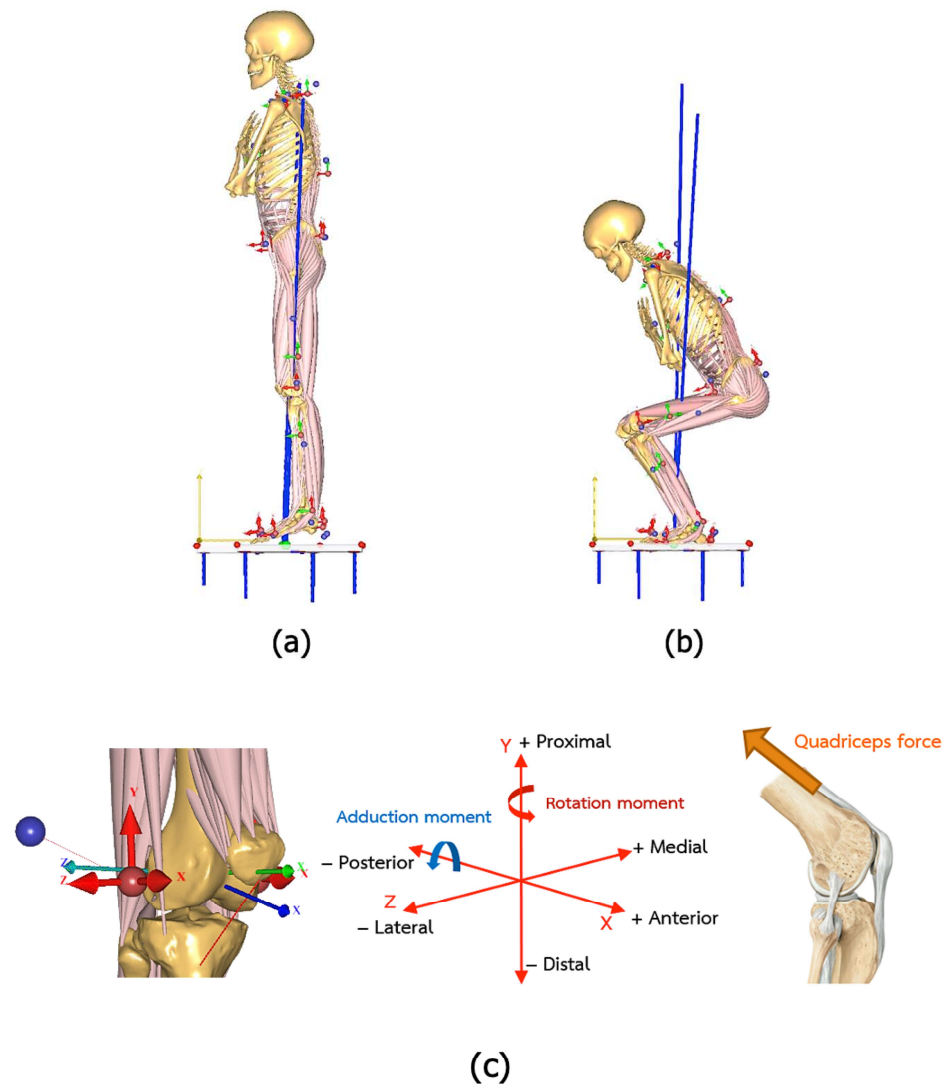


Figure 3. Musculoskeletal model used for inverse dynamics analysis: (a) standing, (b) squatting, and (c) joint coordinate systems. In panel (c), the orange arrow represents the quadriceps muscle force vector. Red, green, and blue arrows define the local coordinate axes (X, Y, and Z), while curved arrows indicate the direction of the rotation and adduction moments.

2.4. Protocol for Squatting Activity

Participants stood barefoot on the force plates with feet parallel and shoulder-width apart, without using external support devices. The experimental protocol consisted of three standardized phases for each trial:

1. Static Phase: The subject stood still for 5 s to record static calibration data.
2. Squatting Phase: The subject performed a controlled descent into their maximum comfortable knee flexion angle, maintaining heel contact with the force plates and using a self-selected trunk inclination. PS-TKA participants were instructed to squat only within a pain-free, safe range of motion.
3. Repetition and Selection: Each participant completed ten squatting trials. The three best trials were selected for analysis based on movement consistency, the absence of compensatory trunk or foot motion, and complete marker visibility.

2.5. Statistical Analysis

Sample size adequacy was assessed using Cochran's formula, assuming 90% confidence and 10% tolerance, which confirmed that 62 knees provided sufficient statistical power [40]. All statistical analyses were performed using MedCalc Statistical Software version 20.009 (MedCalc Software Ltd., Ostend, Belgium).

Data are reported as mean $\pm$ standard deviation (SD). Differences in kinetic parameters between the two main groups (Healthy vs. PS-TKA) were assessed using independent *t*-tests. To evaluate the effect of body mass, differences across the three BMI categories (Light, Middle, Heavy) within each group were analyzed using one-way analysis of variance (ANOVA). Statistical significance was defined as $p < 0.05$. This study was reported in accordance with the STROBE (Strengthening the Reporting of Observational Studies in Epidemiology) guidelines for cross-sectional studies.

3. Results

3.1. Flexion Angle Characteristics and Kinematic Profiles

The kinematic analysis revealed fundamental disparities in squatting strategies between the native healthy knees and the PS-TKA group, as illustrated in the flexion profiles in Figure 4.

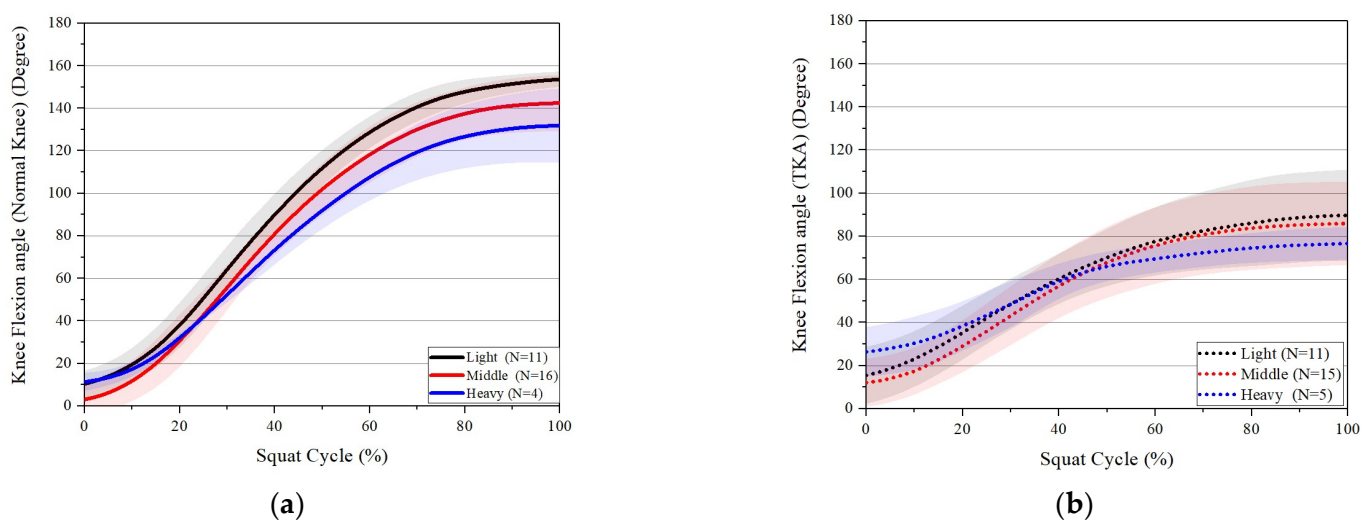


Figure 4. Knee flexion angle profiles during squatting for (a) native healthy knees and (b) PS-TKA knees across light, middle, and heavy BMI categories. The solid and dotted lines represent the mean values, while the shaded areas (color shadows) indicate the standard deviation (SD) for each group.

In the native healthy group (Figure 4a), participants exhibited a smooth, continuous, and deep descent. The flexion trajectories were characterized by a rapid initial flexion phase followed by a steady increase to maximum depth. A clear inverse relationship between BMI and maximum flexion angle was observed. The Light-BMI subgroup demonstrated superior mobility, frequently achieving hyper-flexion angles exceeding 140° and, in some cases, approaching 150° (full deep squat). The Middle-BMI subgroup consistently reached approximately 120–130°, while the Heavy-BMI subgroup, although more restricted, still achieved flexion angles beyond 100°. The shaded standard deviation bands indicate a moderate degree of kinematic variability, reflecting individual differences in natural squatting styles and soft tissue flexibility.

Conversely, the PS-TKA group (Figure 4b) demonstrated a markedly flattened and constrained kinematic profile. Across BMI categories, the peak flexion angles plateaued consistently below 80–90°. The trajectories for the Light, Middle, and Heavy BMI sub-

groups were tightly clustered, showing minimal separation compared to the healthy cohort. This uniformity suggests that the kinematic pathway in TKA patients is predominantly determined by the prosthesis's mechanical hard stops—specifically, the cam-post engagement mechanism—rather than by the patient's physiological potential or body mass. Due to this significant discrepancy in range of motion (ROM), the subsequent kinetic comparative analysis was standardized to a common functional arc of 25° to 75° to ensure valid inter-group statistical comparisons.

3.2. Comparative Analysis of Kinetic Parameters (Healthy vs. PS-TKA)

The comparative analysis of joint kinetics, normalized to body weight (BW) and height (BW·H), highlighted significant mechanical alterations post-arthroplasty (Figure 5). A summary of the peak kinetic parameters comparing Native Healthy Knees and PS-TKA Knees is presented in Table 2. Native healthy knees consistently exhibited greater force and moment generation ($p < 0.05$) throughout the flexion cycle.

- **Anterior–Posterior (AP) Shear Force:** Healthy knees generated significantly higher AP shear forces, with peak differences reaching approximately 0.55 BW compared to PS-TKA knees (Figure 5a). The healthy knee profiles showed a progressive increase in AP force with flexion depth, reflecting the tensioning of the cruciate ligaments (or their functional equivalents in native mechanics) to stabilize the joint. In contrast, PS-TKA knees showed blunted AP forces, likely due to the inherent stability provided by the posterior-stabilized design, which limits physiological rollback and shear translation.
- **Medial–Lateral (ML) Shear Force:** A similar trend was observed in ML forces, where healthy knees exhibited values approximately 0.45 BW higher than TKA knees, particularly in the Light and Middle BMI groups (Figure 5b). The lower ML forces in the TKA group may indicate a simplified, hinge-like motion, restricted by the congruency of the polyethylene insert, thereby reducing the natural varus-valgus toggle observed in native knees.
- **Proximal–Distal (PD) Compressive Force:** The most pronounced difference was found in the vertical compressive loading. Native knees sustained substantial PD forces, averaging around 3.0 BW and peaking at higher flexion angles (Figure 5c). In sharp contrast, PS-TKA knees exhibited significantly lower compressive loads, ranging from 0.45 to 1.0 BW across the analyzed arc. This substantial reduction suggests that TKA patients may subconsciously offload the operated limb or that the restricted flexion prevents the joint from entering the high-compression deep-flexion zone.
- **Joint Moments:** The resultant joint moments further corroborated the force data. The Adduction Moment, a proxy for medial compartment loading, peaked at 5.07% BW × height in healthy knees, whereas TKA knees showed significantly lower values (Figure 5f). Similarly, the Rotational Moment in healthy knees reached 1.29% BW × height, reflecting the natural screw-home mechanism and rotational freedom. In contrast, TKA knees exhibited minimal rotational torque (Figure 5e), consistent with the implant's constrained rotational freedom.
- **Quadriceps Muscle Demand:** The computed quadriceps muscle forces were significantly higher in the healthy group, averaging 4.5 BW, compared to approximately 4.0 BW in the PS-TKA group (Figure 5h). While the absolute difference (~0.5 BW) appears modest, the consistent statistical significance across flexion intervals suggests a persistent deficit in extensor mechanism activation in patients with TKA, potentially indicative of quadriceps avoidance patterns or residual inhibition.

Table 2. Summary of maximum kinetic parameters (Mean $\pm$ SD) comparing Native Healthy Knees and PS-TKA Knees during squatting.

Parameter	Native Healthy Knees (<i>n</i> = 31)	PS-TKA Knees (<i>n</i> = 31)	<i>p</i> -Value
Max. Flexion Angle ($^{\circ}$)	135.4 $\pm$ 12.5	82.3 $\pm$ 8.6	<0.001
Max. Resultant Force (BW)	3.50 $\pm$ 1.00	2.90 $\pm$ 1.20	<0.05
Max. AP Shear Force (BW)	3.52 $\pm$ 0.85	2.91 $\pm$ 0.75	<0.05
Max. Quadriceps Force (BW)	4.50 $\pm$ 0.95	4.05 $\pm$ 0.88	<0.05
Max. Adduction Moment (%BW $\times$ height)	5.07 $\pm$ 1.15	2.36 $\pm$ 0.92	<0.001
Max. Rotation Moment (%BW $\times$ height)	1.29 $\pm$ 0.45	0.85 $\pm$ 0.30	<0.05

3.3. Impact of BMI on Kinetics in Native Healthy Knees

Figure 6 presents a detailed breakdown of kinetic parameters for healthy native knees stratified by BMI. The analysis confirms that body mass is a primary driver of joint kinetics in the natural knee.

- **Load Escalation with Mass:** A direct proportionality was observed between BMI and kinetic magnitude. The Heavy-BMI subgroup consistently exhibited the highest absolute values for AP, ML, and PD forces, as well as resultant moments.
- **Deep Flexion Dynamics:** The Light-BMI subgroup, capable of the deepest flexion (up to 150 $^{\circ}$), demonstrated a unique loading profile where forces were distributed over a larger range of motion. In contrast, the Heavy-BMI subgroup showed sharper increases in Quadriceps Force (Figure 6h) and Resultant Force (Figure 6d) even at lower flexion angles (60–90 $^{\circ}$).
- **Quadriceps Intensity:** The demand on the quadriceps muscle increased markedly with BMI. The heavy group required significantly higher extensor forces to control the descent against gravity, with peaks occurring between 90 $^{\circ}$ and 120 $^{\circ}$ of flexion. This underscores the high muscular demand required to stabilize a heavier body mass during deep squatting.

3.4. Kinetic Invariance Across BMI Categories in PS-TKA Knees

Figure 7 illustrates the kinetic results for the PS-TKA group, revealing a critical and unexpected finding: the dissociation between BMI and kinetic variability.

- **Mechanically Constrained Forces:** Resultant forces and moments were generally low and uniform. AP forces averaged 2.9 BW, whereas ML forces were negligible (0.04 BW), resulting in overlapping standard deviations across all BMI groups (Figure 7a,b).
- **Uniform Loading Profiles:** Unlike the native knees, the kinetic curves for Light, Middle, and Heavy BMI groups in the TKA cohort were remarkably similar. For instance, Quadriceps Forces averaged approximately 4.0 BW, with no statistically significant differences among BMI subgroups (Figure 7h).
- **Implication of Findings:** The lack of BMI-dependent modulation in the TKA group strongly suggests that the mechanical constraints of the prosthesis dominate the joint's kinetic behavior. The standardized geometry of the femoral and tibial components, combined with the cam-post mechanism, appears to constrain movement in a rigid pattern that overrides the natural influence of body mass. Additionally, a restricted range of motion (<90 $^{\circ}$) prevents heavier patients from achieving deep flexion angles, where the effect of body mass on joint loading is most pronounced. This mechanical masking effect is a key differentiator between prosthetic and native knee biomechanics.

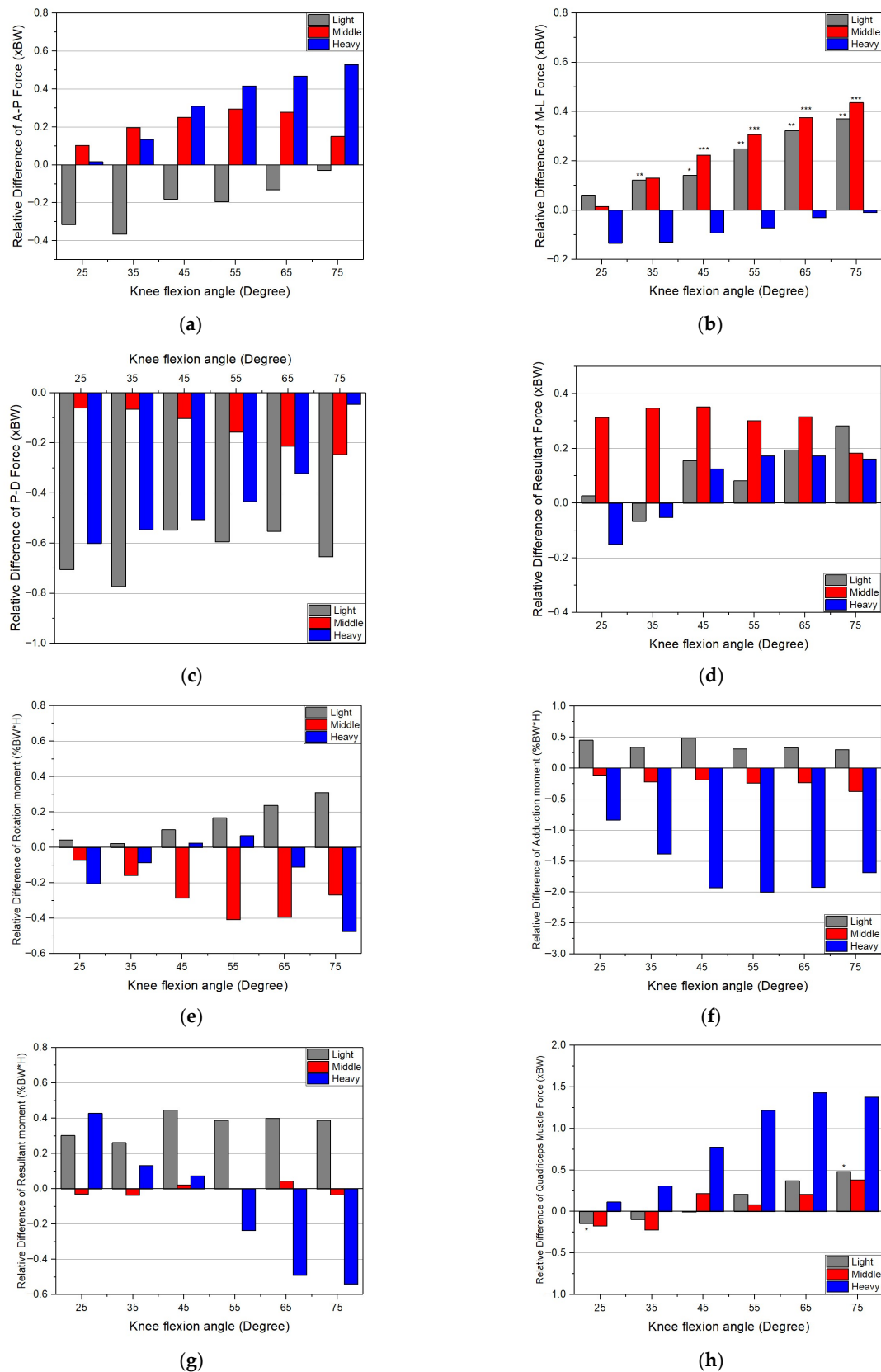


Figure 5. Comparison of knee kinetic parameters between healthy and PS-TKA knees: (a) anterior–posterior force, (b) medial–lateral force, (c) proximal–distal force, (d) resultant force, (e) rotation moment, (f) adduction moment, (g) resultant moment, and (h) quadriceps force. BW = body weight; BW × H = body weight × height. * $p < 0.05$, ** $p < 0.01$, and *** $p < 0.001$ indicate statistical significance levels.

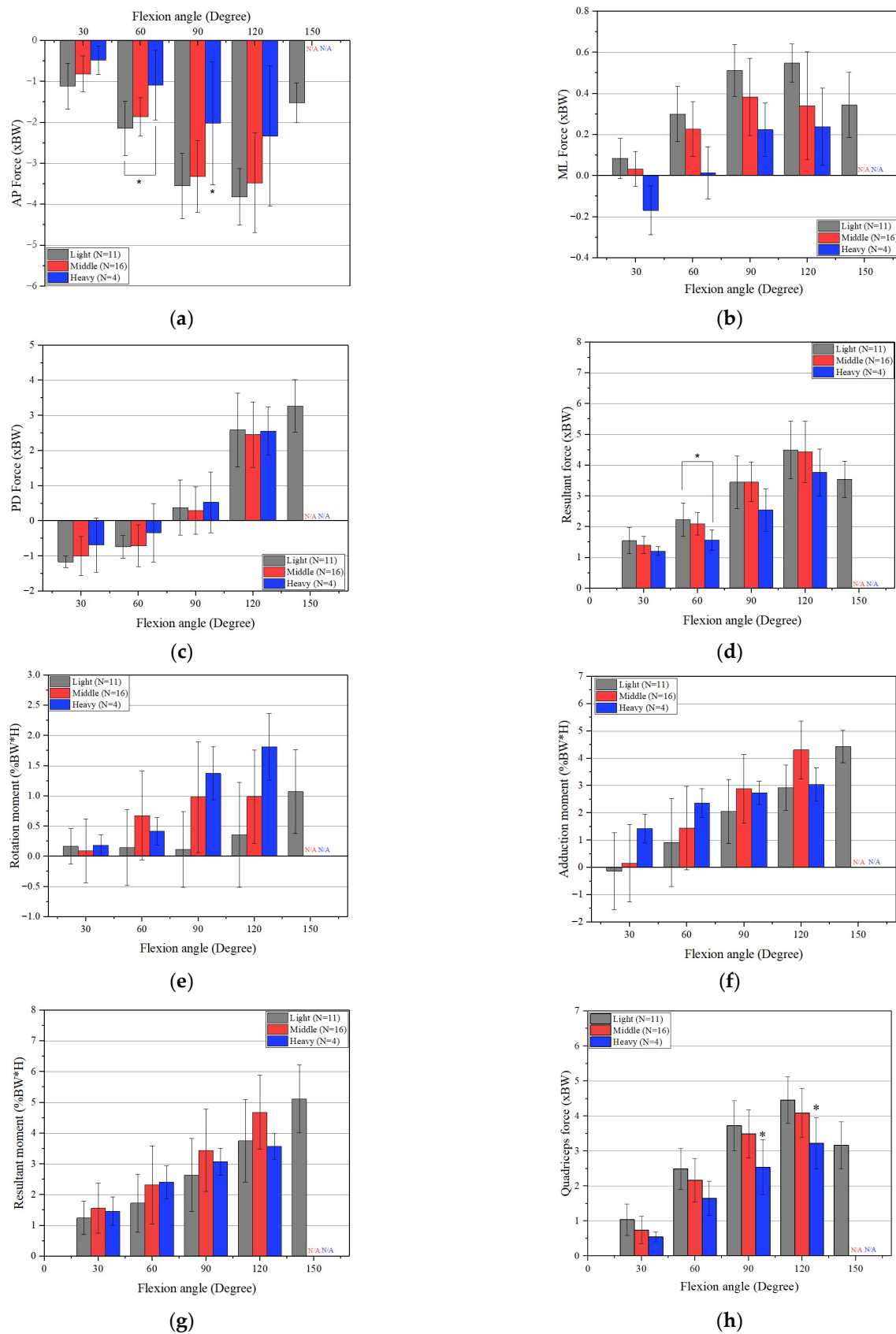


Figure 6. Knee kinetics of healthy knees across BMI categories: (a) AP force, (b) ML force, (c) PD force, (d) resultant force, (e) rotation moment, (f) adduction moment, (g) resultant moment, and (h) quadriceps force. BW = body weight; BW $\times$ H = body weight $\times$ height; N/A = not available. * $p < 0.05$, indicate statistical significance levels.

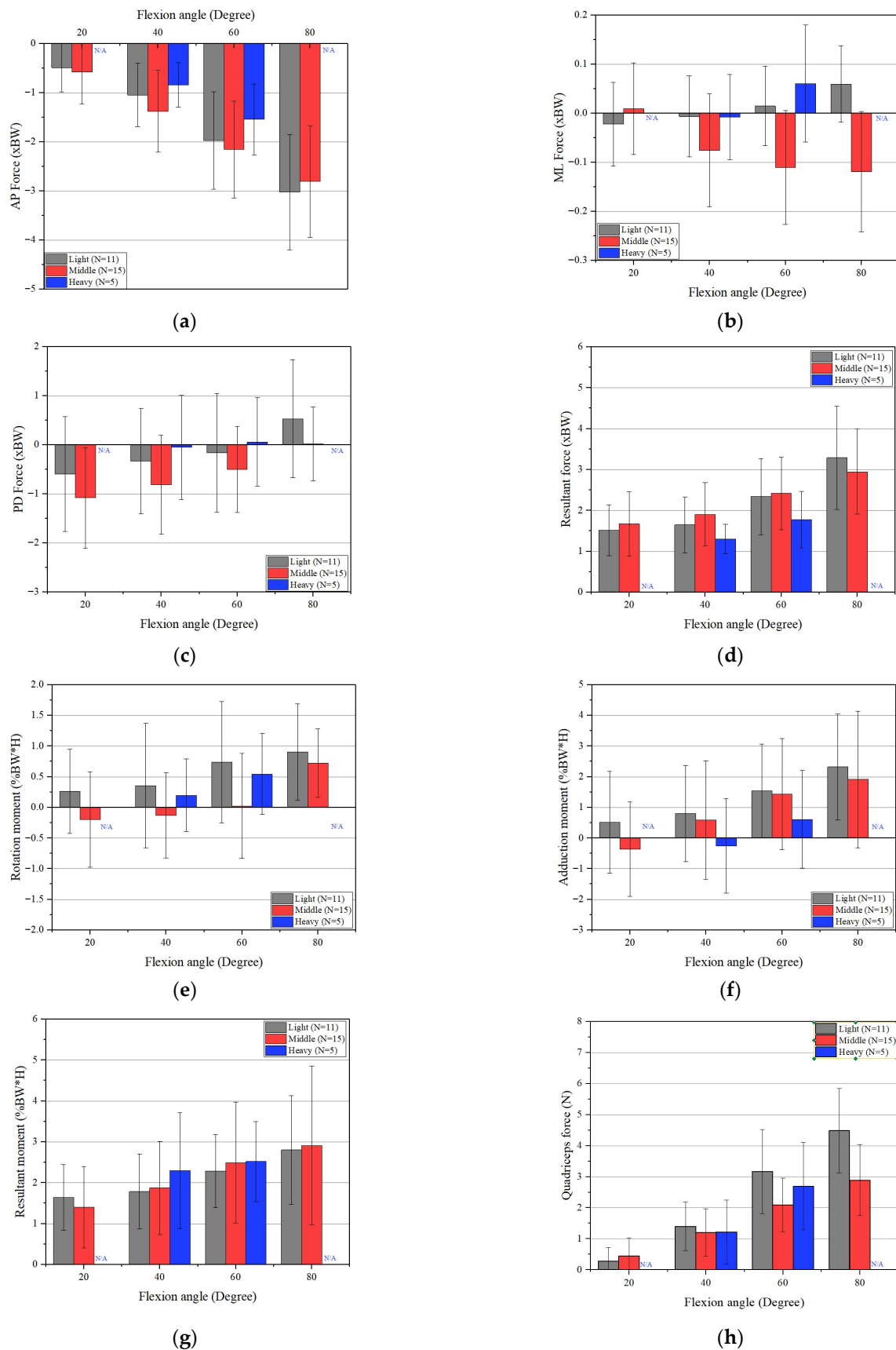


Figure 7. Knee kinetics of PS-TKA knees across BMI categories: (a) AP force, (b) ML force, (c) PD force, (d) resultant force, (e) rotation moment, (f) adduction moment, (g) resultant moment, and (h) quadriceps force. BW = body weight; BW × H = body weight × height; N/A = not available.

4. Discussion

The present study investigated knee joint kinetics during squatting by comparing native healthy knees with posterior-stabilized total knee arthroplasty (PS-TKA) using 3D motion capture and inverse dynamics. The results demonstrated substantial differences in joint loading patterns, flexion capability, and quadriceps demand between the two knee conditions, with additional modulation by body mass index (BMI). The magnitude and trends of knee joint forces observed in healthy knees were consistent with previous reports, where squatting loads in non-arthritic subjects typically ranged between 2.8 and 3.8 body weights (BW) at flexion angles beyond 120° [38], and tibiofemoral forces were reported between 4.7 and 5.6 BW in high-flexion tasks [14]. Studies of non-standard or constrained squatting also documented reaction forces up to 7.8 BW at 70–90° of flexion [41], supporting the high loading demands of deep squatting. This high-load environment in native knees is physiologically managed by the intact menisci and ligaments, which assist in load distribution and stability. However, resection of these structures during TKA fundamentally alters the load-transmission pathway.

In contrast, the resultant forces observed in PS-TKA knees in this study aligned with earlier findings, typically ranging from 1.8 to 3.2 BW in both in vivo instrumented implants and computational analyses [20,31,42]. The lower peak forces in PS-TKA than in native knees may partly result from an inability to achieve the deep flexion angles at which the highest compressive loads typically occur. The peak flexion angles measured in this study showed that maximum flexion exceeded 140° in healthy knees and remained below 80° in PS-TKA knees—comparable to previous research on Asian squatting, where healthy subjects frequently reached 125–146° [9,12], and PS-TKA patients ranged between 80 and 105° depending on implant type and analysis method [31,42]. This large disparity in achievable flexion reflects differences in soft-tissue compliance, posterior condylar rollback, and prosthesis geometry. Surgical factors, including alignment, ligament balance, and implant design, further influence postoperative motion [43], explaining the restricted flexion in the PS-TKA group. Specifically, the PS design relies on the cam-post mechanism to induce femoral rollback. While this provides stability, it may not perfectly replicate the complex rolling-gliding kinematics of the native knee required for deep flexion (>130°), creating a mechanical limit that patients cannot voluntarily overcome.

A key finding of the present work is the influence of BMI on knee loading patterns. In healthy native knees, higher BMI was associated with increased compressive forces, greater quadriceps demand, and elevated joint moments, particularly at greater flexion angles. These trends agree with previous studies linking obesity to changes in gait mechanics, increased joint loading, and accelerated osteoarthritic progression [21,43,44]. Mechanistically, increased body mass increases inertial forces, requiring the quadriceps to generate greater extensor moments to control the descent phase of squatting. Obesity has also been associated with altered postural strategies; for example, obese individuals typically exhibit reduced trunk flexion during sit-to-stand, which lowers knee extensor moments while increasing muscular effort elsewhere [28]. A similar adaptation may explain some of the altered moment patterns observed in the high-BMI group in this study.

A critical finding of this study is the dissociation between BMI and joint kinetics in the PS-TKA group, in contrast to the strong correlation observed in native knees. This phenomenon is likely attributable to the inherent mechanical constraints of the Posterior-Stabilized (PS) prosthesis. The cam-post mechanism in PS-TKA dictates a specific kinematic pathway, thereby restricting the kinematic variability that typically arises from increased body mass in native knees. Essentially, the implant's mechanical constraints override the natural influence of body mass, enforcing a standardized movement pattern regardless of the patient's weight. Furthermore, TKA patients often adopt a 'protective guarding

strategy' due to reduced proprioceptive feedback and laxity of the soft-tissue envelope. This conscious or subconscious stiffening of the joint—characterized by potential co-contraction of the hamstring and quadriceps—limits the compensatory range of motion typically observed in individuals with high BMI, thereby masking the dynamic influence of body mass. Furthermore, in patients with high BMI, limited flexion range necessitates soft-tissue approximation. In obese individuals, physical impingement between the posterior thigh and calf acts as a mechanical barrier, restricting deep flexion regardless of the implant's theoretical range of motion. This soft-tissue contact, which can provide load-sharing in native deep squatting, effectively becomes an obstruction in TKA patients who cannot achieve the angles required to utilize this support mechanism.

The use of 3D motion capture and inverse dynamics in this study follows approaches widely applied in biomechanical research [1,14,19,27,28,45]. These methods provide reliable estimates of joint reaction forces and muscle demands, although limitations inherent to marker-based systems must be acknowledged. Retro-reflective markers are susceptible to occlusion, soft-tissue motion artifacts, and inconsistent placement, which can introduce kinematic noise and influence subsequent inverse dynamics calculations. More advanced methods, such as fluoroscopy or 2D–3D registration, can reduce these uncertainties [46] but are less feasible for routine functional assessments.

The kinetic data produced here also had implications for prosthesis design and rehabilitation. Knee contact forces and loading profiles are essential for predicting tibial insert wear, component stresses, and long-term mechanical behavior [47]. Excessive or uneven load distribution is known to accelerate wear, loosening, and implant failure, and biomechanical factors, including alignment, conformity, and joint kinematics, strongly influence prosthesis longevity [31,48,49]. The finding that TKA loading patterns are relatively rigid and unresponsive to BMI suggests that current PS designs prioritize stability over kinematic adaptability. The population-specific kinetic information derived from this study provides valuable insight into loading patterns during deep squatting—a culturally relevant activity in many Asian populations. The reduced flexion and altered loading in PS-TKA may partially explain functional limitations in Asian populations that frequently perform deep-flexion activities and may support implant design or evaluation tailored to high-flexion requirements. Future rehabilitation strategies should focus on compensatory muscle strengthening to accommodate these altered loading patterns, specifically targeting quadriceps control to manage the abrupt loading changes that may occur when the cam-post engages.

Several limitations must be acknowledged. First, there is a significant age disparity between the healthy control group (mean age ~25 years) and the TKA group (mean age ~67 years). The healthy group was selected to represent a 'Biomechanical Gold Standard' or an optimal functional target, rather than to serve as direct physiological control. Readers should interpret the results as a comparison between postoperative function and ideal native mechanics, while accounting for age-related sarcopenia as a confounding factor. Second, due to the functional limitations of some TKA participants, the dynamic analysis was restricted to a flexion range of 25–75°. While this ensures data consistency, it may not capture peak loads occurring at deep flexion (>90°). Third, this study focused on the operated knee; future research should analyze the contralateral non-operated knee to assess compensatory asymmetry. Finally, squatting speed was self-selected to reflect natural movement, which may introduce variability in inertial forces. Additionally, only posterior-stabilized TKA implants were examined, and other implant designs may exhibit different kinematic and kinetic profiles. Including multiple implant types would provide a broader understanding of design-dependent biomechanical performance.

5. Conclusions

In conclusion, this study confirms that, whereas native healthy knees exhibit load modulation proportional to BMI, PS-TKA knees exhibit a constrained kinetic profile that is independent of patient weight. This suggests that the cam-post mechanism and soft-tissue limitations in PS-TKA override the natural influence of body mass on joint kinetics. Clinically, this implies that current PS-TKA designs provide stable but mechanically rigid kinematics that do not replicate the flexible load-sharing mechanisms of the native knee, particularly in high-flexion activities required by Asian populations. Rehabilitation strategies should therefore focus on compensatory muscle strengthening to accommodate these altered loading patterns.

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Institutional Review Board Statement: The study was conducted in accordance with the Declaration of Helsinki and was approved by the Institutional Review Board (or Ethics Committee) of Suranaree University (Approval No.: 97/2561, 12 February 2020).

Informed Consent Statement: Informed consent was obtained from all subjects involved in the study.

Data Availability Statement: The original contributions presented in this study are included in the article. Further inquiries regarding the datasets can be directed to Supakit Rooppakhun at supakit@sut.ac.th, as the raw data contain sensitive participant information and are subject to ethical and legal restrictions.

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