

Article

Discordant Ranking of Contact Force and Friction Moment Across Activities of Daily Living: A Reanalysis of Instrumented Hip Implant Data

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Abstract

Procedures that predict or test wear in total hip replacements use the contact force waveform as their principal input. This approach rests on the assumption that activities producing large contact forces also impose a large frictional burden. Because friction is governed by sliding at the articulating surfaces and by the state of the lubricating film, the correspondence between the two quantities requires verification. The present study reanalysed publicly available measurements from an instrumented hip implant to test whether the ranking of peak contact force across activities agrees with the ranking of peak friction moment, and to establish how that result depends on how the metric is defined. Hip Joint III data from the OrthoLoad database were used. Of the activities performed by ten subjects, the eight with a cyclic structure were analysed, comprising 72 individual trials; one-legged stance was treated separately because the timing of its peaks is not defined. The accuracy of the data import was confirmed by comparing the imported HIGH100 standardised loads with the originally published values. At the activity level, an exhaustive permutation test over all $8! = 40,320$ orderings was applied; at the trial level, a linear mixed model including the activity-by-metric interaction was fitted. The rankings of contact force and friction moment diverged at the peak (Spearman rho = 0.67; mean rank shift 1.25). In the trial-level model, the activity-by-metric interaction was significant (Wald chi-square = 40.3, df = 7, $p = 1 \times 10^{-6}$) and was largest for jogging (-1.74 ± 0.43 standard deviations, $p = 0.0001$). When the same data were expressed as cycle impulse, the rankings agreed almost completely (rho = 0.98; mean rank shift 0.25) and the interaction was not significant (chi-square = 1.7, $p = 0.975$). The instant at which the friction moment peaked differed by activity: in walking, stair ascent and descent, and jogging it occurred 29 to 37% of the cycle after the contact force peak, whereas in postural transitions and cycling the two instants coincided. The discordance between contact force and friction moment is confined to peak values and does not appear in cumulative quantities. The delayed friction moment peak is consistent with a within-cycle change in lubrication regime, in which the film thins under sustained load and the greatest interfacial shear arrives after the greatest load. Given that polyethylene wear depends on the shear at the articulating surfaces, a load waveform alone may not reproduce the phase of the cycle in which wear actually occurs.



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Keywords: total hip replacement; instrumented implant; friction moment; contact force; in vivo loading; cross-shear; mixed-effects model; circular statistics

1. Introduction

Total hip replacement is among the most successful of orthopaedic procedures, yet revision may become necessary, and osteolysis induced by wear particles is one of its principal causes [1,2]. Research aimed at predicting wear and improving implant design has continued for decades, and most of it takes joint contact force as the input. ISO 14242-1, which specifies simulator testing, prescribes the contact force waveform of the gait cycle together with flexion and rotation angles [3], and numerical wear predictions rely on the same input [4–6].

This approach assumes that activities producing large contact forces also impose a large frictional burden. Wear, however, is determined not by contact force itself but by the sliding that occurs at the articulating surfaces and by the state of the lubricating film between them [7–9]. Whether contact force is an adequate surrogate for friction therefore remains a question to be examined separately.

Answering it requires measurements of both quantities in the same joint at the same time. Instrumented implants satisfy this requirement [10,11], and their measurements are available through a public database, so that reanalysis is possible. Bergmann et al. [12] used these data to define standardised loads for nine activities and reported contact forces and friction moments together. In their discussion, they noted that the peak friction moment during jogging was 6% lower than during walking and 9% lower than during stance, and that the friction moment did not always reach its maximum at the same instant as the contact force. They did not, however, compare the rankings of the two quantities statistically, and the difference in timing was mentioned only for stair descent.

Subsequent work on friction moments has followed two directions. One has tracked the time-resolved friction coefficient and its variation in detail, concentrating on walking as a single activity [13–15]. The other has compared the absolute magnitude of the friction moment with the fixation limit of the cup in order to identify hazardous activities; Bergmann et al. [16] measured approximately 33,000 friction moments during more than 1400 activities and reported that 124 of them reached or exceeded a limit of about 4 N·m, with a maximum of 11.5 N·m. That study addressed the magnitude of the friction moment; its ranking relative to contact force was not part of the analysis.

A comparison of the two rankings across activities, and a quantification of the relationship between the instants at which the two quantities peak, have therefore not been carried out. The present study addresses three questions. The first is whether the ranking of peak contact force across activities agrees with the ranking of peak friction moment. The second is whether any disagreement is confined to peak values or persists when the cumulative burden over the cycle is considered. The third is whether the friction moment peaks at the same instant as the contact force. Answering these questions provides a basis for judging under which conditions the load input used in wear prediction and testing is adequate and under which conditions it requires supplementation.

2. Background

2.1. *In Vivo* Load Measurement with Instrumented Implants

Direct measurement of joint loads *in vivo* was established by the instrumented implants of Bergmann and colleagues. The first two hip implant designs (Types I and II) carried three strain gauges and measured only the three components of contact force, so they provided no information about friction [17,18]. The measuring principle was subsequently enhanced to record all six load components, three forces and three moments, and this six-component design was introduced chronologically in the instrumented vertebral body replacement [19], the shoulder [20], and the knee [21–23] before it reached the hip with the Type III implant. Hip Joint III houses six semiconductor strain gauges and a

nine-channel telemetry transmitter within a hollow neck and measures three force and three moment components simultaneously [10,11]. It is this design that makes it possible to compare contact force and friction moment in the same joint at the same instant.

The measurements were released through the OrthoLoad database. Because the magnitude and time course of the waveforms varied considerably between subjects, a time-warping procedure was developed to obtain representative patterns [24], and applying it yielded the standardised loads AVER75 and HIGH100 for nine activities [12]. Most load waveforms used in current wear prediction derive from this source [25]. Studies that compared model-derived loads with measured values have shown that models do not fully reproduce inter-individual differences [26,27], and estimation by machine learning has recently been attempted [28].

2.2. Lubrication and Friction Moment in Hip Replacements

An articulation formed by a metal or ceramic head against a polyethylene cup does not sustain a full fluid film and operates in the mixed lubrication regime [8,29]. Under these conditions, the friction moment M is approximated by the product of the contact force F , the head radius r , and the apparent coefficient of friction μ .

$$M(t) \approx \mu(t) \cdot F(t) \cdot r$$

This relation defines the logic of the present test. If μ were constant across activities, M would be proportional to F , and the two rankings would necessarily agree. A discordance in ranking would therefore constitute evidence that μ varies with activity.

A note on this relation is required. It is used here in its simplest point-contact form, applied to the resultant magnitudes M_{res} and F_{res} with the nominal head radius. The friction coefficients reported in the original publications and accompanying the published data set were derived from the measured force and moment components by the procedure described there [12,13], and the two approaches are not numerically identical. The quantity μ in the present study is therefore an apparent, lumped coefficient that absorbs the contact geometry and the distribution of contact pressure, and it is used only for comparison between activities within the same implant. Any constant factor cancels in a comparison of rankings, and the primary quantities compared in this study are the measured M and F themselves; the conclusions do not depend on how μ is normalised.

In vivo measurements have repeatedly shown that μ is not constant. It varies substantially within a single step [13], changes with postoperative time [14], and rises by 32 to 143% on average, and by up to 621% in individual subjects, when walking is resumed after a short rest [15]. The last finding was attributed to synovial fluid being squeezed out of the contact during the period of rest, indicating that friction depends less on the magnitude of the load than on the state of the film at that instant. Measurements showing that friction is converted into heat and raises the temperature of the articulating surfaces point in the same direction [30–32]. The influence of load and body weight on friction under walking conditions has also been demonstrated in simulator studies [33].

The regime in which a bearing operates is set by the ratio of the lubricant film thickness to the composite roughness of the two surfaces. Where the film is much thicker than the roughness (fluid-film lubrication), the surfaces are separated, and friction arises from viscous shear of the fluid, so it depends on sliding velocity and viscosity and only weakly on load. Where the film is thinner than the roughness (boundary lubrication), load is carried by asperity contact and adsorbed protein layers, and friction is approximately proportional to load. Between these limits (mixed lubrication), both mechanisms act, and μ falls as velocity rises, the relation described by the Stribeck curve [5,9]. Elastohydrodynamic estimates place a hard head against polyethylene in the mixed regime for most of the gait

cycle, with film thicknesses of the order of the surface roughness [8]. Even hard-on-hard bearings, which sustain thicker films, move between regimes within a cycle and between activities: for the ISO gait cycle, the calculated conditions are predominantly fluid-film, whereas stair descent brings up to about a third of the cycle into boundary lubrication [34]. A further element is time. The film is squeezed out under sustained load and replenished when the load is released, so that friction depends on the load history of the preceding fraction of a second as well as on the instantaneous load and velocity [15,35]. Friction in a joint replacement is therefore expected to vary with activity rather than to be a constant of the material pair, and this expectation is the premise of the comparison between activities made here.

2.3. When Wear Occurs: The Role of Cross-Shear

A second consideration concerning the use of contact force as the sole input arises from the mechanism of wear itself. The wear equation of Archard [7] makes wear volume proportional to the product of load and sliding distance, but for polyethylene this form does not account for experimental observations. Wear rates increase markedly under multidirectional motion, in which the sliding direction changes continuously, compared with unidirectional sliding [36]; the phenomenon was explained in terms of molecular orientation [37]; subsequent experiments quantified the dependence of wear rate on crossing angle [38]; and computational models improved when cross-shear and contact pressure were incorporated alongside load and sliding distance [39].

The implication of these studies is that wear proceeds not at the instant of maximum load but at the instant of maximum shear at the articulating surfaces. The friction moment is a direct measure of that shear, so the instant at which it peaks bears directly on the input conditions of wear models. No report quantifying the relationship between the two peak instants across activities was identified.

A qualification is needed for the polyethylene analysed here. The cross-shear sensitivity established in these studies refers to conventional UHMWPE. In highly cross-linked polyethylene (XPE), the cross-linked network restricts molecular reorientation, and the wear factor is markedly lower and considerably less sensitive to the cross-shear ratio, although it does not become independent of it [40]. For the 32 mm ceramic-on-XPE bearing of the Hip Joint III implant, cross-shear is therefore a diminished rather than an absent factor. The argument of this study does not, however, rest on cross-shear alone: the friction moment is a direct measure of interfacial shear whatever the wear law, so the instant at which it peaks is relevant to any model in which wear is driven by sliding under load, and most relevant where cross-shear amplifies that dependence. This point is taken up in Section 5.5.

2.4. Load Inputs in Wear Testing

ISO 14242-1 adopts the gait cycle as a single reference [3]. This choice serves reproducibility and comparability of testing, but it carries two assumptions. The first is that walking represents the frictional burden. Patients, however, do not confine themselves to walking [41,42], and the magnitude and direction of loading differ between activities [18,43]. The second assumption is that a single standardised waveform represents the patient population. Since activity level itself varies widely between individuals and wear is a function of use rather than of time [44,45], the position that a single representative value occupies within the distribution needs to be established. Reports that simulator design alters the sliding track and hence the wear outcome [46,47], together with evidence that the size distribution of wear particles governs the biological response [48,49], indicate that the validity of the load input is linked to the reliability of the test result as a whole.

2.5. What Earlier Work Established, and What This Study Addresses

Table 1 summarises what previous studies established and the scope of the present work. The study that defined the standardised loads reported both quantities but did not compare their rankings; the studies that examined friction in detail were confined to walking; and the study that addressed the magnitude of the friction moment focused on comparison with the fixation limit of the cup. The present study compares the two rankings across activities, examines how the result depends on the definition of the metric, and quantifies the phase relationship between the two peak instants for each activity.

Table 1. What earlier studies established and the scope of the present work.

Study	Data and Scope	What Was Reported	What Was Not Addressed	Scope of the Present Study
Bergmann et al. [12]	10 subjects, nine activities, standardised loads	AVER75 and HIGH100 defined; peak values and time courses of F and M. The discussion notes that M during jogging is 6% lower than during walking and that the M peak does not coincide with the F peak	The rankings of the two quantities were not compared statistically; the difference in timing was given only for stair descent	Rankings compared across eight activities using an exhaustive permutation test and a mixed-effects model
Bergmann et al. [16]	Nine subjects, >1400 activities, ~33,000 measurements	Peak M compared with a cup fixation limit of about 4 N·m; 124 activities reached or exceeded it, maximum 11.5 N·m	The magnitude of M was the object; its ranking relative to F was not analysed	The ranking relationship between F and M and the phase of the peak instants, rather than absolute magnitude
Damm et al. [13–15]	Walking only	Time-resolved friction coefficient, postoperative change, and a rise of up to 621% after rest	Comparison between activities was outside the scope	Extension to eight activities with a test of differences between them
Bender & Bergmann [24]	Time-warping method	A procedure for deriving representative waveforms from strongly varying signals	The position of the derived representative value within the individual distribution was not examined	Percentile position of the standardised load computed from individual data
Bragdon et al. [36]; Wang [37]; Kang et al. [39]	Specimens and computational models	Cross-shear governs polyethylene wear	The instant of maximum shear was not measured in vivo	Phase of the friction moment peak obtained from in vivo data

This work provides three kinds of evidence. First, it establishes from measured data whether contact force can serve as a surrogate for the frictional burden and, if so, under which definition of the metric. As long as wear prediction and testing take the contact force waveform as their input, this question bears directly on the validity of the input condition. Second, the phase of the friction moment peak is information required by wear models that incorporate cross-shear [39], so the activity-specific phase values reported here can serve as input to such models. Third, knowing where the standardised load lies within the distribution of individual subjects allows the choice of load level in a test condition to be justified explicitly, and provides a starting point for weighting the composition of activities when combined with activity frequency data [41,44].

3. Methods

3.1. Data

The complete Hip Joint III standardised load data set provided by the OrthoLoad public database (<https://orthoload.com/database/>, accessed on 4 August 2026) was used [50]. The data were derived from the measurements of Bergmann et al. [12] and can be down-

loaded without registration. Following the data availability statement of the original study, the records were retrieved by selecting Hip Joint III as the implant and querying by activity. The data were downloaded on 4 August 2026. The construction and specification of the implant are described in the device documentation of the database (<https://orthoload.com/hip-joint-iii/>, accessed on 4 August 2026). The implant consists of a TiAl6V4 stem with a 32 mm alumina ceramic head articulating against a highly cross-linked polyethylene liner, so the head radius used in the calculation of the friction coefficient is $r = 0.016$ m. The ten subjects (Table 2) are designated H1L to H10R; loads are reported in a right-femur coordinate system, and data from left implants are mirrored to the right [51].

Table 2. Subjects and data.

Item	Value
Subjects	10 (H1L to H10R)
Body weight	90.5 ± 13.4 kg (range 77.5 to 122.1)
Implant	Hip Joint III, 32 mm Al_2O_3 head, XPE liner
Activities measured	9
Activities in the main analysis	8 (one-legged stance excluded)
Trials in the main analysis	72 (7 to 10 per activity)
Samples per trial	601 (cycle normalised to 0 to 100%)

The raw data are provided in three units for each trial: measured newtons, normalised to 75 kg, and normalised to body weight. All three were imported so that the influence of the normalisation on the results could be examined. Importing 82 individual trials and nine averaged-subject activities yielded 282 trials and 169,482 sample points. Because the raw files use comma decimal separators and CRLF line endings, both were converted during parsing. The main analysis used the individual data in measured newtons.

3.2. Verification of the Data Import

To confirm the accuracy of the parser and the unit conversion, the imported peak values of the HIGH100 standardised loads were compared activity by activity with the values published by Bergmann et al. [12]. Across all nine activities, the mean absolute deviation was 0.0% for contact force and 0.2% for friction moment. This confirms that the differences observed in the analyses that follow do not arise from errors in the import.

3.3. Definition of the Metrics

For each trial, the maximum, the 95th percentile, and the cycle impulse of the resultant force F_{res} and the resultant moment M_{res} were computed. The 95th percentile was included because a maximum may be determined by a single sample. The phase within the cycle at which each quantity peaked was also computed, together with the difference between them, defined as the phase of the M peak minus the phase of the F peak. The apparent friction coefficient was calculated as the ratio of $M_{\text{res}}(t)$ to the product of $F_{\text{res}}(t)$ and r , restricted to the interval in which F_{res} was at least 10% of $F_{\text{res,max}}$ because the ratio diverges as the denominator approaches zero at low loads.

3.4. Selection of Activities and Treatment of One-Legged Stance

Of the nine measured activities, the eight remaining after the exclusion of one-legged stance were taken as the main analysis set. One-legged stance is a steady-state postural task involving repeated small adjustments for balance, and Bergmann et al. [12] reported that its moment components varied excessively and irregularly and that the peaks in the three planes did not occur at the same instant. In the present data, the between-trial concentration of the friction moment peak phase was $R = 0.38$, against a mean of 0.69 for the other activities, and the concentration of the contact force peak phase was likewise the

lowest of all activities. In this activity, the instant of the peak is therefore not defined, and phase-based metrics cannot be applied on the same footing as for the other activities. The reason for the exclusion is the difference in the nature of the task rather than the reliability of the measurements, and the results for this activity are presented separately in Section 4.7.

3.5. Statistical Analysis

At the activity level, Spearman rho between the ranking of F and the ranking of M across the eight activities was computed. Because the sample is small, exact p -values were obtained by enumerating all $8! = 40,320$ permutations rather than relying on an asymptotic approximation. Since the null hypothesis of a rank correlation test is that of no association, the magnitude of the discordance was reported separately: bootstrap resampling ($B = 10,000$, random seed 20260804) provided a 95% confidence interval for the mean rank shift [52].

At the trial level, F and M were standardised separately and arranged in long format, and a linear mixed model was fitted with the main effects of activity and metric and their interaction as fixed effects and subject as a random intercept. In this model, the activity-by-metric interaction expresses the fact that the profile across activities differs between the two metrics, and a joint Wald test over all interaction terms served as the primary test.

Because the phase difference is a circular quantity, it was converted to an angle, and the mean direction, the resultant vector length R , and the Rayleigh test were computed for each activity [53–55]. Differences between activities were assessed by a label permutation test, since the assumption of equal concentration required by the Watson–Williams test may not hold.

Four procedures were used to examine the stability of the conclusions: refitting with each subject omitted in turn, refitting under the three normalisations of load, refitting with the aggregation rule changed from the maximum to the 95th percentile, and recomputing μ for head diameters of 28, 32, and 36 mm.

3.6. Reproducibility

All analytical steps were carried out with the order of execution and the computing environment recorded, and the values reported here can be recomputed by the same procedure. The analyses used Python 3.9 with NumPy 2.0.2, SciPy 1.13.1, pandas 2.3.3, and statsmodels 0.14.6 [56–59].

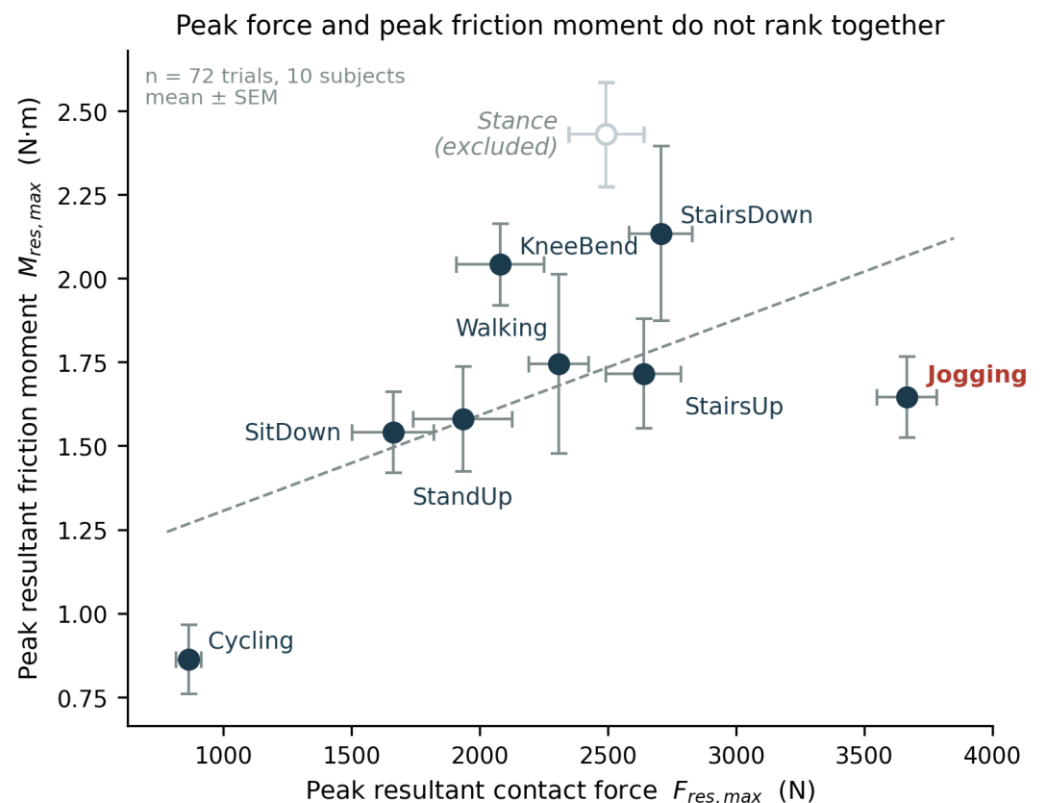
4. Results

4.1. Contact Force and Friction Moment by Activity

Table 3 gives the descriptive statistics for the eight activities. Peak contact force was largest during jogging and smallest during cycling, reflecting the mechanical demand of each activity. The ranking of peak friction moment differed. It was largest during stair descent, followed by knee bending, whereas jogging, which produced the largest contact force, ranked fifth of the eight activities (Figure 1).

Table 3. Descriptive statistics for the eight activities (individual subjects, measured in newtons).

Activity	<i>n</i>	Peak Value		Peak Phase (% of Cycle)			Apparent Friction Coefficient μ		Cycle (s)
		$F_{\max}$ (N)	$M_{\max}$ (N·m)	F	M	$\Delta\phi$	$\mu_{\max}$	μ_{mean}	
Jogging	7	3666	1.65	40	72	+32	0.148	0.045	0.74
Stair descent	9	2705	2.13	26	59	+32	0.155	0.065	1.49
Stair ascent	9	2639	1.72	30	59	+29	0.120	0.058	1.51
Walking	10	2307	1.75	25	53	+28	0.135	0.052	1.13
Knee bending	9	2079	2.04	65	43	−21	0.117	0.060	5.44
Standing up	9	1934	1.58	40	44	+4	0.098	0.050	3.20
Sitting down	9	1662	1.54	47	47	0	0.095	0.053	3.63
Cycling	10	862	0.86	24	23	0	0.099	0.061	1.45

**Figure 1.** Peak contact force and peak friction moment by activity (mean ± SEM). Although the two quantities increase together overall, the rankings of individual activities do not agree. The dashed line is the least-squares fit. One-legged stance, excluded from the main analysis, is shown in grey.

Values are means across the n trials of each activity. n , number of trials; $F_{\max}$, peak resultant contact force; $M_{\max}$, peak resultant friction moment; F phase and M phase, instant within the normalised cycle (% of cycle) at which F_{res} and M_{res} reach their maxima; $\Delta\phi$, phase of the M peak minus phase of the F peak (positive values mean that M peaks later); $\mu_{\max}$, largest instantaneous apparent friction coefficient; μ_{mean} , cycle-mean apparent friction coefficient; both were computed as $M_{\text{res}}/(F_{\text{res}} \cdot r)$ with $r = 0.016$ m over the interval in which F_{res} was at least 10% of $F_{\text{res,max}}$. Cycle, mean cycle duration.

4.2. Discordance at the Peak

At the activity level, the rank correlation for peak values was $\rho = 0.67$, with an exact permutation p -value of 0.083. Because this p -value refers to the hypothesis of no association, the magnitude of the discordance is reported separately: the bootstrap gave a mean rank shift of 1.25 positions with a 95% confidence interval of 1.00 to 2.25.

In the trial-level mixed model, the activity-by-metric interaction was significant (joint Wald chi-square = 40.3, $df = 7$, $p = 1 \times 10^{-6}$). The largest individual coefficient was that of jogging (-1.74 ± 0.43 standard deviations, $p = 0.0001$), meaning that, relative to walking, the friction moment during jogging was 1.74 standard deviations lower than would be expected from the contact force. In the opposite direction, knee bending rose from fourth in contact force to seventh in friction moment (Figure 2, left).

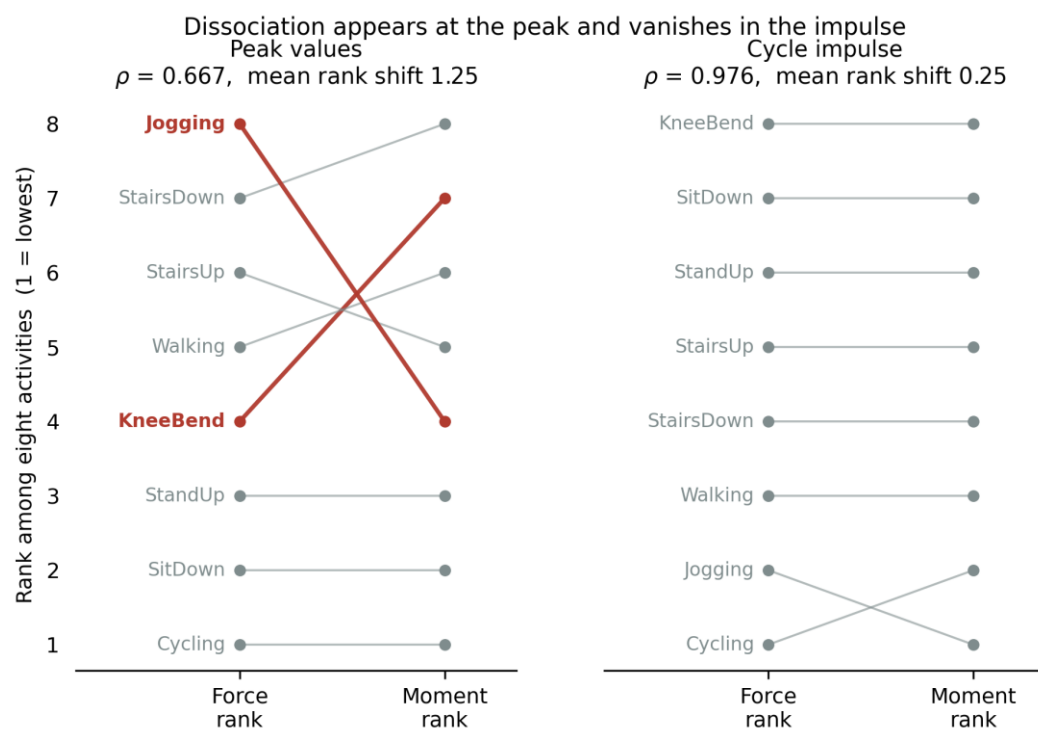


Figure 2. Movement from the contact force ranking to the friction moment ranking. Left, peak values; right, cycle impulse. At the peak, jogging and knee bending move in opposite directions, whereas in the impulse almost every activity retains its position. Bold lines mark activities that move by three or more positions.

4.3. Agreement in the Cycle Impulse

The result changes when the same data are expressed as cycle impulse. The rank correlation was $\rho = 0.98$, and the mean rank shift was 0.25 positions (95% confidence interval 0.25 to 0.75). The interaction in the mixed model was not significant (chi-square = 1.7, $df = 7$, $p = 0.975$). The mean rank shift of 1.25 positions for peak values against 0.25 for the impulse is a fivefold difference, and the two confidence intervals do not overlap (Table 4, Figure 2).

Table 4. Ranking comparison by definition of the metric.

Metric	Spearman Rho	Exact p	Mean Rank Shift	95% CI	Wald Chi-Square (df = 7)	p
Peak value	0.67	0.083	1.25	1.00 to 2.25	40.3	1×10^{-6}
95th percentile	0.69	0.069	1.00	1.00 to 2.25	35.9	8×10^{-6}
Cycle impulse	0.98	0.0004	0.25	0.25 to 0.75	1.7	0.975

Exact p -values are from the enumeration of all $8! = 40,320$ permutations, with the null hypothesis of no association. Mean rank shifts and confidence intervals are from a bootstrap resampling of subjects ($B = 10,000$).

4.4. The Instant of the Peak

The phase difference in the friction moment peak separated the activities into two groups (Table 5, Figure 3). In cycling, sitting down, and standing up, the difference concentrated near zero, with resultant vector lengths between 0.56 and 0.86. In walking, stair ascent, stair descent, and jogging, it clustered between +29 and +37%. Knee bending had $R = 0.28$ and no defined direction. The difference between activities was significant in a label permutation test carried out over all nine measured activities ($p = 0.0001$).

Table 5. Phase difference in the friction moment peak.

Group	Activity	<i>n</i>	Mean $\Delta\phi$ (%)	<i>R</i>	Rayleigh <i>p</i>
Simultaneous	Standing up	9	+4	0.86	0.0002
Simultaneous	Cycling	10	−1	0.83	0.0002
Simultaneous	Sitting down	9	+1	0.56	0.056
Delayed	Stair descent	9	+37	0.63	0.025
Delayed	Jogging	7	+37	0.58	0.093
Delayed	Stair ascent	9	+32	0.45	0.157
Delayed	Walking	10	+29	0.69	0.006
Undefined	Knee bending	9	−18	0.28	0.519

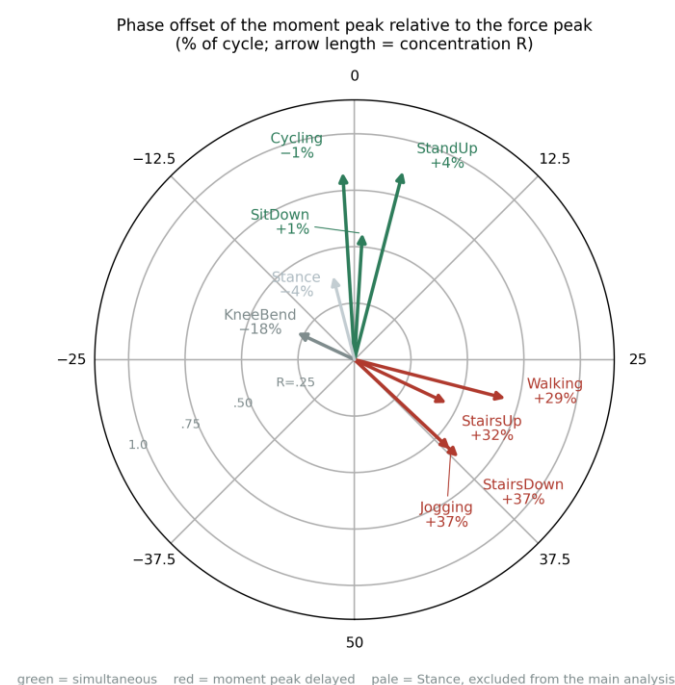


Figure 3. Phase difference in the friction moment peak shown in polar coordinates. The direction of each arrow is the mean phase difference, and its length is the concentration *R*. In activities involving ground contact impact, the friction moment peaks occur about 30% of the cycle after the contact force; in postural transitions and cycling, the two peaks coincide. The interpretation of these two groups in terms of lubrication regime is given in Section 5.2.

4.5. Robustness

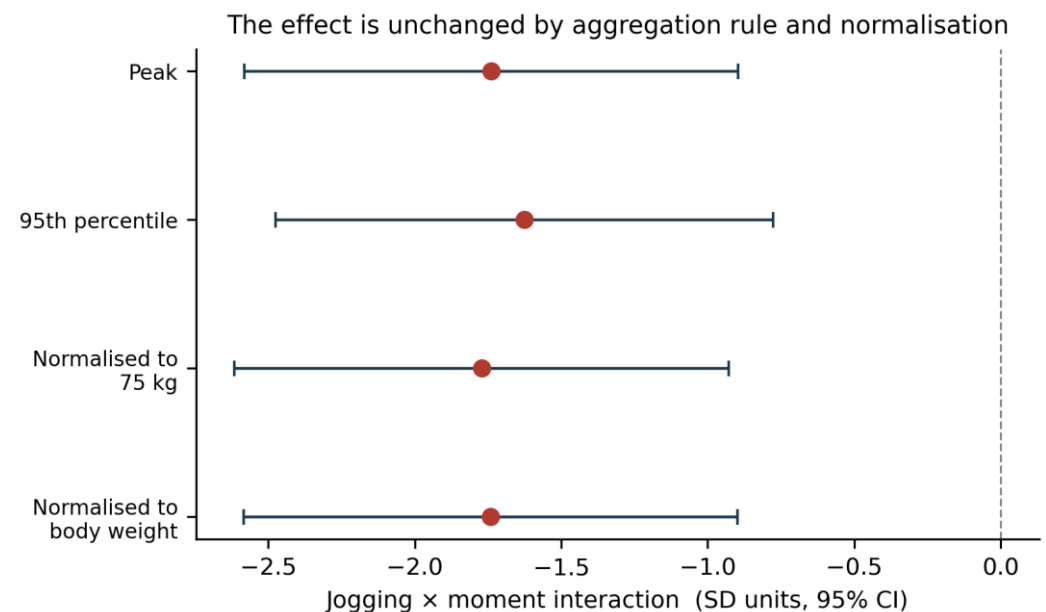
In the eight refits with one subject omitted in turn, the largest *p*-value of the joint interaction test was 1.0×10^{-4} , and the coefficient for jogging lay between −1.33 and −2.01. The influence of individual subjects, which is the first concern with a small sample, was not observed.

The results were unchanged under the three normalisations of load (Table 6). Chi-square ranged from 40.3 to 41.4 and the jogging coefficient from −1.74 to −1.77. Given that body weight ranged from 77.5 to 122.1 kg, the agreement indicates that the discordance in ranking does not arise from differences in body weight.

Table 6. Refitting under three normalisations of load.

Normalisation	Wald Chi-Square (df = 7)	<i>p</i>	Jogging Coefficient (SD)
Measured newtons	40.3	1×10^{-6}	-1.74 ± 0.43
Normalised to 75 kg	41.4	1×10^{-6}	-1.77 ± 0.43
Normalised to body weight	41.0	1×10^{-6}	-1.74 ± 0.43

When the aggregation rule was changed from the maximum to the 95th percentile, the interaction remained significant (chi-square = 35.9, $p = 8 \times 10^{-6}$) with a jogging coefficient of -1.63 ± 0.43 (Figure 4). The sensitivity to head diameter is determined algebraically: since μ is the ratio of M to the product of F and r , a change in r scales all activities by the same factor and the ranking is unaffected. The cycle-mean μ ranged from 0.052 to 0.074 for a 28 mm head, from 0.045 to 0.065 for 32 mm, and from 0.040 to 0.057 for 36 mm.

**Figure 4.** Jogging interaction coefficient with 95% confidence intervals under different aggregation rules and normalisations. The estimates and intervals are nearly identical across the four conditions.

4.6. Position of the Standardised Load Within the Individual Distribution

The percentile occupied by the standardised load AVER75 within the distribution of individual peak contact forces ranged from 30 to 67% depending on the activity (Table 7, Figure 5). Walking was the lowest at the 30th percentile, and sitting down and standing up were the highest at the 67th. When the ranking analysis was applied to the standardised load data themselves, the correlation for peak values was $\rho = 0.45$ for AVER75 and 0.33 for HIGH100, lower than for the individual data, while for the impulse it was 0.98 and 0.95, as high as for the individual data.

Table 7. Percentile position of the standardised load AVER75.

Activity	<i>n</i>	AVER75 (N)	Percentile
Walking	10	1859	30
Cycling	10	724	40
Jogging	7	3055	43
Knee bending	9	1702	44
Stair ascent	9	2222	56
Stair descent	9	2272	56
Sitting down	9	1351	67
Standing up	9	1582	67

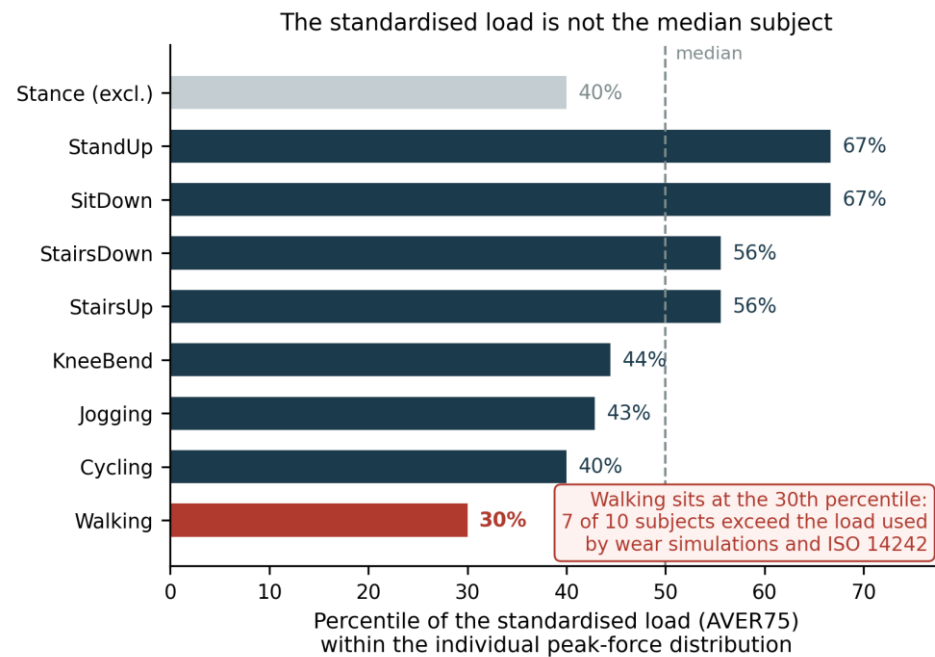


Figure 5. Position of the standardised load within the distribution of individual subjects. The dashed line marks the median. The walking load, which wear testing and numerical prediction use most often, occupies the lowest position.

Percentiles were computed from the data normalised to 75 kg.

4.7. One-Legged Stance

For one-legged stance, the peak contact force was 2494 N and the peak friction moment 2.43 N·m, the largest friction moment among the nine activities. This value, however, depended strongly on the aggregation rule. The difference between the mean of the per-trial maxima and the maximum of the mean waveform was 26%, against a mean of 15% for the eight activities, and the deviation from the standardised load AVER75 was 53%, against a mean of 7% for the eight activities.

In this activity, components oscillating more than 30 times per cycle accounted for 6% of the total variance of the friction moment signal, compared with 0.4% in the other activities. In the contact force of the same trials, this fraction was 0.09%, no different from the other activities: the oscillation appeared in the friction moment alone. The peak persisted over 13 consecutive samples above 90% of its maximum, and at the instant of the peak the three moment components stood at an average of 67% of their own maxima. The ratio of the largest to the median peak across subjects was 1.48, so the value was not driven by particular individuals.

These findings indicate that the observed friction moment is not measurement noise. The between-trial concentration of the peak instant was nonetheless low at $R = 0.38$, so that the magnitude of the peak is real while its timing is not fixed within the cycle. Because of this property, phase-based metrics cannot be applied to this activity in the same way as to the others, and it was therefore separated from the main analysis.

5. Discussion

5.1. Principal Findings

Reanalysis of instrumented hip implant data across eight activities established three findings. The rankings of contact force and friction moment diverged at the peak (Figure 1 and the left panel of Figure 2); when the same data were expressed as cycle impulse, the rankings agreed almost completely (Figure 2, right; Table 4); and the instant of the friction

moment peak either coincided with the contact force peak or followed it by about 30% of the cycle, depending on the type of activity (Table 5, Figure 3).

That the two peaks occur at different instants means that the moment of greatest load and the moment of greatest shear at the articulating surfaces are separated. A single contact force waveform cannot represent both instants, and because the separation varies from 0 to 37% between activities (Table 5), it cannot be handled by a single correction either. Given that polyethylene wear proceeds in phases of high shear [36,37], a procedure driven by the load waveform alone may fail to reproduce the phase in which wear actually occurs. In activities such as postural transitions and cycling, where the two instants coincide, this problem does not arise, so the representativeness of a load waveform depends on which activity is chosen as the test condition.

5.2. Mechanism of the Discordance

From the relation between M , μ , F , and r given in Section 2.2, a discordance in ranking implies that μ varies with activity. In the present data, the cycle-mean μ ranged from 0.045 to 0.065 and the instantaneous maximum from 0.095 to 0.155 (Table 3). This variation is ordered along the mechanical character of the activities, and the following paragraphs interpret it in terms of the lubrication regime, which in a hard-on-polyethylene bearing lies in the mixed region between boundary and fluid-film conditions (Section 2.2).

That jogging had the largest negative interaction coefficient in the present study can be explained by the velocity dependence of mixed lubrication. Previous work indicates that a higher relative velocity at the articulating surfaces produces a thicker film, reduces solid contact, and lowers the friction coefficient [8,29]. Jogging is the activity with the highest relative velocity among the eight examined here, and it also has the shortest cycle (0.74 s) with a pronounced swing phase in which the load falls to a small fraction of its peak; both conditions favour replenishment of the film before the next impact. In simulator experiments, a lower swing-phase load reduces friction in the following stance phase [33], and in vivo the friction coefficient rises sharply when the film has been squeezed out during a period without motion [15]. The finding that the friction moment of jogging ranked in the middle despite the largest contact force (Table 3) is consistent with this mechanism, and appears in Figure 1 as the position of jogging below and to the right of the fitted line. In the opposite direction, knee bending in the present study combined a moderate contact force with a high friction moment (rising four positions in Figure 2, left); this activity has a long cycle of 5.44 s and a large range of joint motion, corresponding to sustained slow sliding over a wide arc under a load that is never released. Under these conditions, the film is thinned by being squeezed out and not replenished; the contact moves towards boundary lubrication, and μ approaches its upper, load-proportional value. The cycle-mean μ of 0.060 for knee bending and of 0.065 for stair descent, against 0.045 for jogging (Table 3), follow this ordering.

The phase results of Figure 3 can be read in the same terms. In the four activities involving ground-contact impact, the contact force rises abruptly at foot strike, at an instant when the joint has just reversed direction, and the entraining velocity is near its minimum. The load is therefore first carried by squeeze-film action of the fluid trapped in the contact, and the friction moment at that instant is comparatively small. As stance proceeds, the load remains high, the film is progressively squeezed out, and the sliding velocity increases as the hip extends, so that solid contact and interfacial shear grow while the load is still substantial. The friction moment peaks 29 to 37% of the cycle later than the force, when the film is thinnest rather than when the load is largest. In postural transitions and cycling, by contrast, load and motion vary slowly and continuously, the entraining velocity is low throughout, and the contact remains close to the boundary regime for the whole cycle; μ is

then approximately constant within the cycle, the friction moment follows the load, and the two peaks coincide ($\Delta\phi$ within $\pm 4\%$, Table 5). Knee bending, with its undefined direction ($R = 0.28$), sits between these cases: its load and motion are both sustained, so no single instant of maximum shear is favoured. The two groups in Figure 3 thus correspond to two temporal patterns of the film, transient squeeze and recovery in impact activities against quasi-steady boundary contact in slow activities, and not merely to two kinematic patterns. This interpretation implies that the instant of greatest wear risk in impact activities is one of depleted lubrication and high sliding velocity, which is the condition under which wear factors measured in the laboratory are highest.

Because the sliding velocity at the articulating surfaces was not measured directly, and the film thickness cannot be measured *in vivo*, this account remains an inference from the friction data and from the established behaviour of lubricated contacts. Cycle duration did not serve as a usable proxy for velocity: the activity with the longest cycle had the lowest cycle-mean μ , showing that cycle length does not reflect the instantaneous sliding velocity at the surfaces. Two further cautions apply. Synovial fluid is a non-Newtonian, protein-containing lubricant whose viscosity and boundary-film properties differ between patients and change after arthroplasty, so the quantitative position of a given activity on the Stribeck curve is not known, and the apparent μ used here lumps the contact geometry into a single coefficient (Section 2.2), so its absolute values should not be compared with laboratory friction factors obtained under defined lubricants.

5.3. Relation to the Observations of Bergmann et al. [12]

Bergmann et al. [12] reported contact forces and friction moments for nine activities together and observed in their discussion that, cycling aside, the peak friction moment during jogging was lower than in all other activities, by 6% relative to walking and 9% relative to stance. The indication of a discordance in ranking was thus already present. The present study converts that indication into a testable form, and the correspondence is set out in Table 1. Specifically, the rankings of the two quantities were compared across eight activities, the limitation of a small activity-level sample was addressed by a trial-level mixed-effects model, and the interaction was found to be significant at chi-square = 40.3, $p = 1 \times 10^{-6}$.

What this test means is that the ordering of values across activities differs between the two metrics. If the friction coefficient were constant across activities, the friction moment would be proportional to the contact force, and the two orderings would agree. The significance of the interaction therefore supports, from the data, the view that the friction coefficient is not a fixed property of the implant but a quantity that varies with activity. The test moreover uses 144 observations from 72 trials rather than eight activity-level points, and so attains a statistical power not available in the activity-level rank correlation. Significance speaks to the existence of a difference and not to its size, so the size is reported alongside it as a mean rank shift of 1.25 positions and a jogging coefficient of -1.74 standard deviations.

The same paper also noted that the friction moment does not always reach its maximum at the same instant as the contact force, but presented this as a remark on a single case, stair descent, and in its tables reported the friction moment at the instant of the peak force and the peak friction moment as separate entries. The present study defines this difference in timing as a circular quantity and computes the mean direction and concentration for each activity, finding a consistent delay of $+29$ to $+37\%$ in the four activities involving ground contact impact (Table 5; the red arrows in Figure 3). For walking, this corresponds to about 0.33 s in a cycle of 1.13 s.

The present study further computes the percentile position that the standardised loads defined by Bergmann et al. [12] occupy within the distribution of individual subjects. The method for deriving representative waveforms was itself presented and validated by Bender and Bergmann [24], but the position of the derived representative value within the distribution was not reported. In the present data, that position ranged from the 30th to the 67th percentile across activities (Table 7, Figure 5).

5.4. Relation to the Body of Work on Friction Moments

Among earlier studies, those that tracked in vivo friction in detail concentrated on walking as a single activity [13–15]. Observing walking alone suggests that contact force and friction moment increase together, but when the present study compared eight activities, the correspondence proved to depend on the activity. The range of activity-specific μ obtained here (Table 3) is consistent with the walking-based reports while showing, through comparison between activities, that μ is not a fixed property of the implant but varies with the activity. This provides grounds for re-examining wear models that assume a single value of μ [5].

Bergmann et al. [16] measured friction moments during more than 1400 activities, identified those exceeding a cup fixation limit of about 4 N·m, and reported a maximum of 11.5 N·m. That study far exceeds the present one in the range of activities and the number of measurements and showed that the absolute magnitude of the friction moment can reach clinically relevant levels. What the present study addresses is not absolute magnitude but the ranking relative to contact force; the two studies answer different questions. The peak friction moments observed here, between 1.5 and 2.1 N·m (Table 3), are smaller than the maximum in that study because the activities examined were limited to the eight standardised ones.

5.5. Peak Timing and Cross-Shear

The implication of the phase results becomes clear when they are connected to the mechanism of polyethylene wear. Wear rate depends not only on load and sliding distance but also on changes in sliding direction [36–38], and computational models improved when cross-shear was included [39]. The results of these earlier studies imply that wear proceeds at the instant of maximum shear at the articulating surfaces rather than at the instant of maximum load.

The friction moment is a direct measure of that shear. In the present study, the friction moment peak in activities involving ground contact impact occurred 29 to 37% of the cycle after the contact force peak (Table 5, Figure 3). A wear model driven by the load waveform alone places the instant of maximum shear at the instant of maximum load, and the resulting error varies from 0 to 37% of the cycle depending on the activity. The activity-specific phase differences and concentrations in Table 5 are in a form that can be supplied directly to models incorporating cross-shear pressure.

The separation into two groups is also mechanically interpretable. In activities with ground contact impact the load rises sharply at the moment the foot strikes the ground while the relative motion of the joint reaches its maximum afterwards, whereas in postural transitions and cycling load and motion vary together and continuously.

Lubrication enters this argument at two points. First, the delay of the friction moment peak is itself a lubrication phenomenon: as set out in Section 5.2, the instant of maximum shear in impact activities is the instant at which the film has been squeezed out under sustained load while sliding velocity is high, that is, the instant at which the contact is closest to boundary conditions. The phase of wear is therefore also the phase of least fluid protection, and a wear model that places maximum shear at maximum load misses not only

the timing of the shear but also the state of the film at which it occurs. Second, laboratory wear factors, including the cross-shear dependence quantified by Kang et al. [39,40], are obtained under a defined serum lubricant and a defined loading waveform; transferring them to the in vivo condition requires that the load, motion, and film state be matched at the same instants of the cycle. The activity-specific phase values in Table 5 supply the timing part of that matching. The fluid part, the thickness and composition of the film at each instant, remains outside the reach of the present data, and the interpretation offered here is accordingly an inference rather than a measurement.

A qualification concerns the polyethylene. Highly cross-linked polyethylene, the liner material of the implant analysed here, is markedly less sensitive to cross-shear than conventional UHMWPE because the cross-linked network resists the molecular reorientation on which the cross-shear mechanism depends; its wear factor is lower and its dependence on the cross-shear ratio flatter, although not absent [40]. Therefore, for the 32 mm ceramic-on-XPE bearing, cross-shear is a secondary rather than a primary determinant of wear, and the expected wear consequence of the phase displacement is small in absolute terms. What the present data establish is the timing of the interfacial shear, not the amount of wear it produces. The timing result holds for any wear mechanism that is driven by sliding under load, cross-shear included, and its practical weight increases with the cross-shear sensitivity of the material: it is most consequential for conventional polyethylene and for wear models of the form proposed by Kang et al. [39], and least consequential for the most highly cross-linked materials. For XPE liners, the more immediate relevance of the friction moment is not wear at all but the frictional torque transmitted to the cup and taper interfaces [16], for which the peak value and its timing are relevant irrespective of the wear law.

5.6. Representativeness of the Load Input

In the present study, the standardised load AVER75 occupied positions from the 30th to the 67th percentile of the individual distribution depending on the activity (Table 7, Figure 5). The walking load, used most often in wear testing and numerical prediction, occupied the lowest position, which means that seven of the ten subjects in these data experienced contact forces above the standardised value.

This result gains weight when set beside work on activity level. Schmalzried et al. [44] showed from clinical data that wear is a function of use rather than of time; Silva et al. [45] reported that patients average about two million walking cycles per year with wide individual variation; and Zahiri et al. [42] argued for the measurement of activity. Those studies addressed the frequency of activities, whereas the present study provides the ranking of the frictional burden across activities and its phase characteristics. Combining the two axes would allow a framework for estimating patient-specific wear exposure.

It would not, however, be appropriate to derive a quantitative correction factor from these data alone because the relation between wear and load is mediated by an activity-dependent μ and by cross-shear, so that linear extrapolation is not justified. Three directions are proposed for testing standards and numerical models. One is to select a specific percentile of the distribution explicitly instead of a single representative waveform. Another is to include activities other than walking with weights derived from activity frequency data [41,45] together with the activity ranking reported here. The third is to specify the phase of the friction moment alongside the load waveform. Efforts in this direction have already been made in part [25].

5.7. Interpretation of One-Legged Stance

In the present study, one-legged stance differed in character from the other eight activities. Its peak friction moment was the largest of the nine activities, while the between-trial concentration of the peak instant was the lowest (shown in grey in Figures 1 and 3). The indices presented in Section 4.7 indicate that this peak is not measurement noise: it persisted over 13 samples, the three moment components were elevated together at the instant of the peak, it was not concentrated in particular subjects, and the high-frequency content appeared in the friction moment but not in the contact force of the same trials.

These observations agree with the account given by Bergmann et al. [12], which described one-legged stance as demanding considerable muscular activity for balance and its moment components as varying excessively and irregularly, while attributing the high friction during stance to synovial fluid being squeezed out under sustained static loading. Taken together, these two statements describe a situation in which small postural adjustments made for balance repeatedly generate friction whose timing is not fixed within the cycle. The peak duration, component synchrony, and frequency characteristics obtained here are consistent with that account.

Two further in vivo observations support this reading. In the hip, Damm et al. [15] found that the friction coefficient during walking rose by 32 to 143% on average, and by up to 621% in individual subjects, when walking was resumed after a short standing rest, and attributed the rise to synovial fluid squeezed out of the contact while the joint was loaded without motion; friction returned to its usual level within a few steps as the film recovered. In the knee, measurements with instrumented tibial components synchronised with fluoroscopy have shown that friction is highly dynamic and phase-dependent within a single activity, that it is only partly linked to the magnitude of the load, and that unloaded phases allow the film to be replenished whereas sustained loading drives the contact towards drier conditions [35]. The one-legged stance is precisely a condition of sustained loading with small, irregular motion, and the large but irregularly timed friction moments observed here are what these two studies would predict for it.

The measurements for this activity are therefore valid, but phase-based metrics cannot be applied to them. The conditions under which friction arises in steady-state postural tasks constitute a separate research question, and activities of long duration require an analytical framework other than cycle normalisation.

5.8. Methodological Considerations

The present study introduced four procedures to secure the reliability of its conclusions while working with data whose sample size cannot be increased. These procedures have not generally been applied in earlier studies of instrumented implant data.

The first is the choice of the unit of analysis. In discussing the difference from their earlier data, Bergmann et al. [12] stated explicitly that the statistical significance of the differences could not be tested because the previous study had included only four subjects. This constraint applies throughout the field of instrumented implant research. Since measurement requires an invasive device, the number of subjects cannot readily be increased, and reporting has consequently remained largely descriptive at the activity level [12,13,17,18]. The present study circumvents the constraint by lowering the unit of analysis from activity to trial. Using 144 observations from 72 trials instead of eight activity-level points, and treating subject as a random intercept, secures a statistical power unattainable from the same data by activity-level rank correlation: for the same data, the activity-level exact test gave $p = 0.083$ whereas the trial-level model gave $p = 1 \times 10^{-6}$ (Table 4). This gain must be interpreted with caution. The number of trials is not the same for every subject (jogging, for example, was available for seven of the ten subjects), so subjects who contributed more

trials carry more weight in the fixed-effect estimates, and the random intercept corrects for the between-subject shift in level but not for this imbalance in weighting. The refits with each subject omitted in turn (Section 4.5) bound the size of the effect: the joint interaction test remained significant in every refit, and the jogging coefficient stayed within -1.33 to -2.01 , so no single subject, however many trials it contributed, determines the conclusion. The precision implied by the trial-level p -value should nevertheless be read as conditional on the ten subjects observed, not as evidence about the population of implant recipients.

The second is the choice of test procedure. To avoid relying on an asymptotic approximation with a small sample, exact p -values were obtained by enumerating all $8! = 40,320$ permutations. Rank statistics are determined by ranks alone, so exhaustive enumeration is possible in principle, and no approximation error is introduced. Since the null hypothesis of a rank correlation test is that of no association, the mean rank shift and its confidence interval were obtained by bootstrap and reported alongside the significance [52]. A test taking perfect agreement as the null hypothesis concerns a value at the boundary of the parameter space and is not valid with a percentile bootstrap, so it was not attempted.

The third is the explicit statement of the aggregation rule and the quantification of its influence. Standardised loads are obtained by time-warping the trial waveforms and taking the maximum of the representative waveform [12,24]. The present study applied both this rule and the averaging of per-trial maxima and reported the difference for each activity. In these data, the difference was 7 to 30% for the friction moment and 0.7 to 15% for the contact force because the friction moment is a signal with greater between-trial phase dispersion. Earlier work presented and validated the procedure for deriving representative waveforms but did not report the differences in value arising from the aggregation rule for each activity. The present study additionally used the 95th percentile, which is insensitive to single samples, to confirm that the conclusions do not depend on the aggregation rule (Table 4, Figure 4).

The fourth is the application of circular statistics to phase data. The difference between the two peak instants is a circular quantity for which the arithmetic mean is not defined. In these data, the arithmetic mean gives a phase difference of +19% for one-legged stance, whereas the circular mean gives -4% , a reversal of sign. In earlier work, the difference in timing between the two peaks was presented as a remark on individual cases or as a separate entry in a table but was not treated with directional statistics [12]. The present study applied the mean direction, the resultant vector length, and the Rayleigh test to all phase metrics [53–55].

In addition, the verification of the data import required of studies that reanalyse public data was carried out. The imported peak values of the standardised loads were compared with the published values, giving mean absolute deviations of 0.0% for contact force and 0.2% for friction moment [12]. This procedure ensures that the differences subsequently observed do not arise from import errors. Refitting with each subject omitted in turn was also used to exclude the influence of individual subjects, following the same principle as the cross-validation schemes used to assess generalisability in small biomechanical data sets [28]. In view of the policy of the group that released the data, stating such verification procedures explicitly is a requirement for reanalysis studies.

5.9. Generalisation to Other Bearing Couples

The results apply directly to a 32 mm alumina head against a highly cross-linked polyethylene liner, the only bearing for which in vivo friction moments have been measured. Whether the discordance between contact force and friction moment would diminish for other bearing couples can be reasoned from the relation $M \approx \mu F r$ and from what is known of their lubrication. The discordance exists because μ differs between activities,

and the ranking of M would coincide with the ranking of F only if μ were the same in every activity. Two limiting regimes make this explicit. Under pure boundary lubrication, μ is approximately constant, M is proportional to F , and the rankings agree. Under full fluid-film lubrication, friction is viscous shear of the film, proportional to velocity and viscosity and only weakly dependent on load, so that M would follow the sliding velocity rather than the load and the discordance would be larger, not smaller. Bearings that operate between these limits show a discordance whose size depends on how far μ moves along the Stribeck curve from one activity to the next.

Hard-on-hard couples such as ceramic-on-ceramic sustain thicker films than polyethylene bearings and have the lowest friction factors under gait conditions [60]. They are not, however, in a single regime: the calculated film thickness for a ceramic-on-ceramic bearing varies several-fold within the ISO gait cycle, and while the ISO cycle is predominantly fluid-film, stair descent brings up to about a third of the cycle into boundary lubrication [34]; their friction factor rises with swing-phase load and depends on the lubricant [60]; and under poor lubrication large hard-on-hard bearings develop friction moments high enough to load the cup fixation and the taper [61]. A bearing that spends part of one activity in the fluid-film regime and part of another in the boundary regime is a bearing whose μ depends strongly on activity. Therefore, for ceramic-on-ceramic, the cycle-mean μ would be expected to be lower than the values in Table 3, but the between-activity variation of μ , and hence the discordance in ranking, would be expected to persist and could be larger because these bearings operate closer to the transition in which μ is most sensitive to velocity and load history. The direction of the individual rank shifts could also differ from those found here: an activity of high velocity such as jogging would be favoured still more by a thick film, whereas a slow, sustained activity such as knee bending, in which the film is squeezed out, would be penalised more. Metal-on-polyethylene bearings, in which a cobalt–chromium head replaces the ceramic head against the same liner, have a film thickness governed mainly by the compliance of the polyethylene, and are expected to behave as the bearing analysed here, with a somewhat higher friction factor for the metal head [60] but the same regime and the same activity dependence.

The two components of the present result should therefore be distinguished when its practical boundaries are drawn. The magnitude of the friction moment and of μ , and the exact positions in the rankings, belong to the bearing measured and should not be transferred to other couples. The phase relationship between the load peak and the motion of the joint is set by the activity, and the delay of the friction moment behind the load in impact activities is expected wherever the film thins during stance, that is, in every couple that is not permanently in the boundary regime; its size in cycle percentage may differ. These expectations are hypotheses, since in vivo friction data exist for one bearing only. They can be tested in simulators by driving different couples with the activity-specific OrthoLoad waveforms and comparing the ranking of measured friction moments with the ranking of applied loads, which is the experiment that the present analysis suggests.

5.10. Limitations

First, there were ten subjects and only seven for jogging, and the number of trials differed between subjects, so that the trial-level estimates weight subjects unequally (Section 5.8). Expanding the sample is not straightforward given the nature of instrumented implant research, but the conclusions were retained on subject-wise refitting, and the estimates agreed under changes in normalisation and aggregation rules (Table 6, Figure 4). Differences in load between subjects are also related to physiological factors such as the degree of muscle damage and cannot be treated as simple variation [62].

Second, sliding velocity and joint angle at the articulating surfaces were not used directly, and the thickness and composition of the synovial film cannot be measured *in vivo*, so the account of the mechanism of the discordance in terms of lubrication regime (Sections 5.2 and 5.5) is an inference. Subsequent analysis combining joint angular velocity data is required.

Third, the apparent friction coefficient is a simplification of spherical contact and does not reflect the stress distribution over the contact area, and it is defined differently from the coefficients reported with the original data set (Section 2.2). Fourth, the results apply to a 32 mm ceramic head against a highly cross-linked polyethylene liner; the expected behaviour of hard-on-hard and metal-on-polyethylene couples is discussed in Section 5.9, but it has not been verified, and generalisation to material combinations with markedly different film thickness requires measurement [34,63]. Fifth, the eight standardised activities do not include atypical activities or extreme loading [16,64]. Sixth, the influence of body weight on friction and complications requires separate consideration [65,66], and the range of body weight in these data is limited.

6. Conclusions

Reanalysis of instrumented hip implant data across eight activities showed that the rankings of contact force and friction moment diverge at the peak and agree in the cycle impulse. This difference means that the adequacy of contact force as a surrogate depends on how the frictional burden is defined.

The instant at which the friction moment peaks separated the activities into two groups. In activities involving ground contact impact, it occurred about 30% of the cycle after the contact force peak, whereas in postural transitions and cycling the two instants coincided. Given that polyethylene wear depends on changes in sliding direction, this suggests that the phase in which wear proceeds may be systematically displaced from the phase indicated by the load waveform.

The standardised loads occupied positions from the 30th to the 67th percentile of the individual distribution, with the walking load used most often in wear testing and numerical prediction at the lowest position. Both the use of contact force as a surrogate for the frictional burden and the use of a single standardised waveform as the input therefore warrant re-examination under these conditions. Total frictional work follows the load, but the conditions at the instant of film breakdown do not; if the latter is what wear testing seeks to reproduce, a load waveform alone is not sufficient.

These results lead to three lines of further work. The first is to establish the activity dependence of the friction coefficient at the level of mechanism. Because sliding velocity at the articulating surfaces was not used here, the account of a shift in lubrication regime remained an inference. Combining the joint angle data provided by the same database to obtain the sliding velocity within the cycle and relating it to the time-resolved friction coefficient would allow a test of whether the differences between activities follow a Stribeck relation [8,13].

The second is to supply the activity-specific phase differences reported here to a wear prediction model and determine whether the outcome changes. Models incorporating cross-shear require the temporal relationship between load and sliding direction [37,39], so comparing the present convention of placing maximum shear at the instant of maximum load with a condition using the phase differences in Table 5 would quantify the size of that difference. Combining activity frequency data [41,45] would then allow a wear prediction weighted by the composition of activities.

The third is to extend the question of representativeness from the magnitude of the load to the distribution as a whole. The present study computed the percentile position

of the standardised load for peak contact force only; the same examination is possible for friction moment and for phase, and the magnitude of between-subject variation is itself a subject for study. Since the present results apply to a 32 mm ceramic head against a highly cross-linked polyethylene liner, studies establishing whether the same discordance appears in hard-on-hard and metal-on-polyethylene couples, for which Section 5.9 sets out the expected direction, are also required [33,34,63].

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Data Availability Statement: The raw data analysed in this study are openly available in the OrthoLoad public database at <https://orthoload.com/database/> (accessed on 4 August 2026). The statistical procedures comprise exhaustive permutation testing of rank correlations, subject-wise bootstrap, mixed-effects modelling, circular statistics, robustness analyses, verification of the data import, and figure generation, with the random seed fixed at 20260804. The analysis scripts are available from the corresponding author on request.

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