Long-term hip loading in unilateral total hip replacement patients is no different between limbs or compared to healthy controls at similar walking speeds

Original Article: Long-Term Hip Loading in Unilateral Total Hip Replacement Patients is no Different between Limbs or Compared to Healthy Controls at Similar Walking Speeds

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Word Count: Excluding references, cover page, abstract & figure captions – 2,993 words
Abstract – 245 words

Keywords: hip contact forces, musculoskeletal modelling, total hip replacement

Abstract

Variation in hip joint contact forces directly influences the performance of total hip replacements (THRs). Measurement and calculation of contact forces in THR patients has been limited by small sample sizes, wide variation in patient and surgical factors, and short-term follow-up. This study hypothesised that, at long-term follow-up, unilateral THR patients have similar calculated hip contact forces compared to controls walking at similar (self-selected) speeds and, in contrast, THR patients walking at slower (self-selected) speeds have reduced hip contact forces. It was further hypothesised that there is no difference in calculated hip contact forces between operated and non-operated limbs at long-term follow-up for both faster and slower patients. Gait analysis data for THR patients walking at faster (walking speed: 1.29 ± 0.12 m/s; $n = 11$) and slower (walking speed: 0.72 ± 0.09 m/s; $n = 11$) speeds were used. Healthy subjects constituted the control group (walking speed: 1.36 ± 0.12 m/s; $n = 10$). Hip contact forces were calculated using static optimisation. There was no significant difference ($p>0.31$) in hip contact forces between faster and control groups. Conversely, force was reduced at heel strike by 19% ($p=0.002$), toe-off by 31% ($p<0.001$) and increased at mid-stance by 15% ($p=0.02$) for the slower group compared to controls. There were no differences between operated and non-operated limbs for the slower group or the faster group, suggesting good biomechanical recovery at long-term follow-up. Loading, at different walking speeds, presented here can improve the relevance of preclinical testing methods.

1 Introduction

Total hip replacement (THR) is a well-established and largely successful orthopaedic procedure, which is primarily employed to treat end-stage osteoarthritis (UK National Joint Registry, 2017). The UK National Joint Registry reports 13-year revision rates of less than 7% for the majority of bearing types (UK National Joint Registry, 2017). Although hip joint contact forces (HCFs) are related to both short-term (e.g. implant micromotion) and long-term THR failure mechanisms (e.g. peri-prosthetic bone remodelling) (Sumner, 2015), direct measurement has only been possible in small samples of disparate patients using instrumented implants (Bergmann et al., 2001; Davy et al., 1988; Rydell, 1966). Musculoskeletal models have been used to calculate HCFs from three dimensional motion capture data and ground reaction forces, with good agreement to instrumented implant measurements (Heller et al., 2001; Modenese et al., 2011; Modenese and Phillips, 2012; Stansfield et al., 2003). Nevertheless, musculoskeletal modelling of HCFs in THR patients has been limited by small sample sizes, short-term follow-up, and wide variation in patient, surgical, and implant factors (Foucher et al., 2009; Li et al., 2014; Stansfield and Nicol, 2002; Wesseling et al., 2016a).

Recently, musculoskeletal modelling of healthy subjects reported that walking speed has a considerable impact on predicted HCFs (Giarmatzis et al., 2017, 2015). Unfortunately, studies reporting HCFs in THR patients have not controlled for walking speed or have been confined to a few patients at short-term follow-up (Bergmann et al., 1993; Heller et al., 2001; Li et al., 2014; Modenese and Phillips, 2012; Stansfield et al., 2003; Wesseling et al., 2016a). It is unclear whether differences observed (e.g. lower HCFs compared to controls) in THR patients relative to healthy subjects are due to walking speed, variable or short-term follow-up, or other surgical outcomes.

Hip moments in THR patients have been reported to be significantly altered at both short-term (Beaulieu et al., 2010) and long-term follow-up (Bennett et al., 2017) compared to controls. At short-term follow-up, differences in HCFs for THR patients compared to controls have also been found across a range of surgical approaches, even with similar walking speeds (Wesseling et al., 2016b). When comparing between limbs, differences have been observed in hip moments between operated and contralateral limbs (Foucher and Wimmer, 2012). However, Li et al. (2014) found no difference in HCFs between limbs 12 months after unilateral THR. To the authors' knowledge, no long-term study has been conducted comparing THR patients' HCFs at different speeds to controls and between limbs.

This study hypothesised that, at long-term follow-up, unilateral THR patients have similar calculated HCFs compared to controls walking at similar (self-selected) speeds and, in contrast, THR patients walking at slower (self-selected) speeds have reduced HCFs. It was further hypothesised that there is no difference in calculated HCFs between operated and non-operated limbs at long-term follow-up for both faster and slower patients.

2 Methods

2.1 Patient and control data

All patients received unilateral THRs in Musgrave Park Hospital (Belfast, UK) under the senior author (DEB) using a posterior approach with the same cemented implants (Orthogenesis custom X-press femoral implant and Elite acetabular implant, DePuy International, Leeds, UK). All had a 28 mm femoral head articulating with an ultra-high molecular weight polyethylene acetabular implant.

Three dimensional gait analysis was undertaken using retroreflective markers placed according to the Helen Hayes marker set (Kadaba et al., 1990). At least three walking trials were performed along a central walkway (8–10m length) and a single representative trial was used for analysis. All patients and control subjects walked bare foot. Data was captured at 120 Hz using a six-camera Vicon 612 system and patients walked at self-selected speed. Patients were stratified by walking speed and the 11 slowest and 11 fastest patients (out of a total of 134 patients) with sufficient force plate data for operated and non-operated limbs were used for musculoskeletal analysis. Force plate data was collected at 120 Hz using two force plates (AMTI, Watertown, MA) and assessed visually using Mokka (Version 0.6, BioMechanical Toolkit) (Barre and Armand, 2014). Force plates were fully concealed and integrated into the floor covering. To avoid targeting, patients were not told of their position and instructed to walk at their normal walking speed. Patients were studied 10 years post-operatively: follow-up ranged from 9.72 to 10.41 years for slower patients and 9.13 to 10.18 years for the faster cohort. Three-dimensional gait data from 10 healthy elderly controls was also used for musculoskeletal analysis. Controls and THR groups' age ranges and body mass index (BMI) are presented in Table 1. Ethical approval was obtained from the local ethics committee (Queen's University Belfast, Faculty of Medicine, Research Ethics Committee, Reference number: 253/02).

2.2 Musculoskeletal modelling

A lower extremity model, “gait2392”, with 12 segments, 23 degrees of freedom and Hill-type models of 92 muscle-tendon compartments was used to analyse gait with a standard static optimisation workflow in OpenSim 3.3 (Stanford University, CA, USA) (Delp et al., 2007, 1990). Lumbar extension, lumbar bending, lumbar rotation, metatarsophalangeal and subtalar joints were fixed in anatomical neutral positions for all analyses. A zero-lag low-pass filter with a cut-off frequency of 6 Hz was applied to ground reaction forces and kinematics (Mantoan et al., 2015). Isotropic scaling of the generic model segments was performed based on markers placed on bony landmarks of the subject with the exception of the pelvis, where the laboratory-measured distance between the left and right anterior superior iliac spines was used (Figure 1).

Inverse kinematics and inverse dynamics were performed to calculate joint angles and moments respectively. Static optimisation was executed to estimate the muscle activations and forces. The objective function for static optimisation minimised the sum of the muscle activations (defined as the ratio of predicted force relative to maximum isometric force for the muscle) squared based on a study showing that this criterion produces realistic HCFs and muscle activation patterns relative to instrumented implant and EMG data (Modenese et al., 2011). In accordance with OpenSim best practice, reserve actuators were added to the joint between the pelvis and the ground to account for dynamic inconsistencies caused by errors in the model estimations of the subject’s inertial parameters and geometry as well as marker placement/tracking inaccuracies. Force-length-velocity relationships were not used for the muscle models in this study as they have been shown not to affect the calculation of HCFs for level walking (Anderson and Pandy, 2001). Finally, a joint reaction analysis was performed in OpenSim in order to calculate the resultant hip force, i.e. HCF. All HCFs were subsequently normalised to body weight (BW) and the moments were divided by body mass in kilograms (Nm/kg).

2.3 Data analysis and statistics

The start and end of the gait cycle were identified and 100-point splines were fitted to the HCFs and hip moments for both sides in MATLAB (2015b, TheMathWorks, Inc., MA, USA). These points were used to create ensemble average curves for the control, slower patient, and faster patient groups. Ensembles curves were also created for comparison between operated and non-operated limbs in both the faster and slower patient groups. Peaks in HCF at heel strike, mid-stance, and toe-off were identified for quantitative statistical analysis. These points within the gait cycle were also used to compare hip adduction moments, while hip flexion moments were compared using maximum and minimum values of a complete gait cycle and hip rotation moment was compared using only the maximum value. Differences between HCFs and moments for operated and non-operated peaks were tested using the paired Wilcoxon Signed Rank-Sum Test as the data were not normally distributed. Control subjects were compared (by hip kinematics, HCFs and hip moments) to the speed-stratified groups with a Wilcoxon Rank-Sum Test using a Bonferroni correction for multiple comparisons. In cases where a Bonferroni correction was used, the p-value has been multiplied by the number of comparisons so that the interpretation is the same for all data, i.e. $p<0.05$ is significant. All statistical analysis was performed using R (Version 3.4.0) (R Core Team, 2017).

3 Results

Maximum hip extension was significantly ($p<0.001$) reduced in slower THR patients compared to controls (Figure 2; Table 2). Similarly, hip adduction range was significantly lower ($p<0.01$) for both THR groups (i.e. slower and faster) compared to controls (Figure 2; Table 2). Overall, hip moments tended to be reduced in the THR groups compared with the control group (Figure 2**;** Table 3). In particular, the slower group had lower peak extension ($p<0.001$; Figure 2) and internal rotation moments ($p=0.002$; Figure 2) when compared to control subjects; this trend was matched by the faster patients for extension ($p=0.002$; Figure 2) and internal rotation moments ($p=0.03$; Figure 2). The abduction moment showed a similar trend to HCFs, i.e. distinct double peak for faster and control groups while slower patients showed a higher mid-stance moment ($p<0.05$), similar in magnitude to the heel-strike and toe-off peaks. In terms of magnitude, the faster THR patients' moments were more comparable to the control group than the slower THR group.

Ensemble averaged HCFs of faster THR patients and control subjects showed a distinct double peak profile whereas slower THR patients exhibited lower HCFs and a plateau during stance phase (Figure 3). There were significant differences in HCFs between slower THR subjects and control subjects at heel-strike ($p=0.002$), mid-stance ($p=0.02$) and toe-off ($p<0.001$) (Table 4 and Figure 3). Conversely, there was no statistical difference ($p>0.31$) between these forces in faster THR patients compared to controls (Figure 3). There were no significant differences between operated and non-operated limb HCFs at heel-strike, mid-stance, or toe-off ($p>0.10$) for either the faster or slower patient groups (Figure 4).

4 Discussion

The first hypothesis of this investigation was that calculated HCFs in THR patients at long-term follow-up are comparable to healthy control subjects at similar walking speeds and HCFs of slower patients are significantly reduced. We also hypothesised that HCFs of operated and non-operated limbs in THR patients were not significantly different at long-term follow-up. The results supported both hypotheses. Slower patients showed a less dynamic HCF profile whereas both faster and control groups showed increased contact forces at the beginning and end of stance phase and lower contact force at mid-stance. There was no difference in hip contact forces throughout the gait cycle between operated and non-operated limbs in either the faster or slower groups.

Hip contact forces from this study compare well with experimentally measured HCFs (via instrumented implants) at similar walking speeds. For example, instrumented implant measurements by Bergmann et al. (1993) at a walking speed of 0.83 m/s (the closest comparison for our slower group speed of 0.72 ± 0.09 m/s) averaged 3.5 BW for peak HCF compared to 2.98 BW in our study. Predicted HCFs, using a similar musculoskeletal modelling approach, for elderly healthy subjects walking at 0.83 m/s also show good agreement with our findings at low walking speeds, peaking at 3.3 BW (Giarmatzis et al., 2017). At a walking speed of 1.38 m/s, and for elderly female patients, Giarmatzis et al. (2017) estimated a mean hip joint loading at the first peak of 4.3 BW and 4.6 BW at the second peak, while Weinhandl et al. (2017) predicted peaks of 3.39 ± 0.45 BW and 4.61 ± 0.55 BW for a cohort of five male and five female young, healthy subjects walking at 1.34 ± 0.07 m/s. The values presented here are similar at 3.5 ± 0.6 BW and 4.5 ± 0.9 BW.

In the studies described above, excluding Weinhandl et al. (2017), subject speed was controlled using a treadmill rather than being self-selected. Similar HCFs have been predicted for disparate cohorts when walking speed was matched, e.g. Giarmatzis et al. (2017) showed very similar force profiles between young and old groups at matched walking speeds between 0.83 m/s and 1.94 m/s. Walking speed in our study was self-selected and patients were not tested on a treadmill. Walking speed is related to physical and mental function (Peel et al., 2013) with faster patients generally expected to have better function. Consequently, our study is likely capturing the best functioning THR patients in the faster cohort and the worst in the slower group.

A previous study (Li et al., 2014) using different musculoskeletal modelling methodology showed similar trends but altered magnitudes: Li et al. (2014) presented HCFs for a relatively young control group (average age: 44.97 years) walking at 1.44 m/s (1.39 –1.50 m/s) and found a considerably lower mean toe-off peak of 3.67 BW compared to controls in this study and the publications by Giarmatzis et al. (2015) and Weinhandl et al. (2017) on healthy subjects. These differences are most likely due to the methodological approaches, with Li et al. (2014) opting for the Anybody software (AnyBody Technology, Aalborg, Denmark) and model while the other studies, including the current, use OpenSim with the “gait2392” model. These two modelling approaches have shown altered muscle force estimations in a comparative study (Trinler et al., 2017), which can be a result of separate scaling and mathematical optimisation processes in each software.

With regard to comparisons between operated and non-operated limbs, the findings in this investigation are in agreement with previous works at short-term follow-up (Beaulieu et al., 2010; Benedetti et al., 2010; Li et al., 2015, 2014). The close matching of hip contact force profiles between operated and non-operated limbs suggests good biomechanical recovery at long-term follow-up. The absence of difference in HCFs between operated and non-operated limbs for both faster and slower patients suggests symmetrical biomechanical recovery is achieved, which is independent of walking speed. Foucher and Wimmer (2012) report an increased abduction moment in the non-operated hip preoperatively compared to speed-matched controls, with no improvement up to 1 year following THR. In contrast, we found no significant difference in hip abduction moment between faster patients compared to normal subjects and significantly lower hip internal rotation and extension moments, which may be due to the longer period of recovery (Figure 2).

Strengths of this study include control of possible confounding surgical and patient factors. In addition, many previous works have assessed HCFs shortly after surgery when patients have variable levels of recovery whereas we report HCFs at long-term follow-up (Bergmann et al., 2001; Heller et al., 2001; Li et al., 2014; Stansfield et al., 2003; Stansfield and Nicol, 2002).

Notwithstanding these strengths, there were several important potential limitations to this study. Firstly, gait kinematics, kinetics, and walking speed have been shown to be age-related (Bennett et al., 2017, 2008), and there was an age difference between the slower THR and control groups in this study. However, Giarmatzis et al. (2017) have shown that walking speed difference would be expected to influence kinetics independent of age. Secondly, patients were stratified by walking speed from a larger dataset (Bennett et al., 2008), meaning patients walking at mid-range speeds were not included. In our opinion, assessing the extremes of walking speed has the advantage of establishing more useful ranges for preclinical testing design that are relevant to the increasing volume of THR procedures being performed (UK National Joint Registry, 2017). Lastly, while the sample sizes in this study are based on previous works with similar aims (Foucher et al., 2009; Li et al., 2014; Stansfield and Nicol, 2002; Wesseling et al., 2016a), future studies involving larger sample sizes would likely add to the findings presented here.

Limitations in the modelling aspects of this investigation include isotropic scaling of a generic musculoskeletal model based on the location of gait markers, as used here and in similar studies (Giarmatzis et al., 2015; Weinhandl et al., 2017). This may be susceptible to misplacing the joint centres and is particularly pertinent since previous validation studies have used CT and X-ray derived joint centres (Modenese et al., 2011; Stansfield et al., 2003). This has increased relevance in THR patients as the location of the hip joint centre is often shifted following THR surgery (Bonnin et al., 2012). Previous studies suggest that the location of the joint centre may be improved through the use of medical imaging (Kainz et al., 2017; Lenaerts et al., 2009) or a functional method (Leardini et al., 1999). Lenaerts et al. (2009) compared CT measurements to a linear scaled generic model, finding an anterior and proximal shift of the hip joint centre location of 30mm and 20mm respectively; this did not cause a significant effect on the magnitude of HCFs but did influence the orientation of the force. Future investigations examining the accuracy of hip joint centre location in THR patients would help in the interpretation of this work.

The findings of this study have important implications for preclinical testing of THR implants and patient recovery. Firstly, there were considerable differences in HCFs (both in force profile shape and magnitude of loading) between slower and faster THR patients suggesting preclinical simulation studies should include this loading variability. Secondly, this study shows that self-selected walking speed is associated with contact force in the hip joint and should therefore influence study design when comparing THR patients to controls. Thirdly, HCFs (and moments, Appendix 1) are similar between operated and non-operated limbs at long-term follow-up suggesting patients have good biomechanical recovery.

In conclusion, the hypothesis that there is no difference between healthy control subject HCFs and THR patient HCFs at similar walking speeds was supported by this study. We also showed there to be large differences between slower THR patients and both control and faster groups. The difference between a faster and slower THR patient, in terms of HCF, is substantial and is relevant to both computational and lab-based preclinical testing methods and many biomechanical failure mechanisms of THR. In addition, we found contact forces to be no different between the operated and non-operated hips at both faster and slower walking speeds. This equalisation may be indicative of the longer recovery period studied here.

Acknowledgements

Nicholas Dunne, Alex Lennon and John O'Connor report grants from Belfast Arthroplasty Research Trust and from Department of Employment and Learning, Northern Ireland. Megan Rutherford reports grants from Belfast Arthroplasty Research Trust related to the current study.

Conflict of interest

The Belfast Arthroplasty Research Trust has received funding from DePuy Synthes and Zimmer Biomet but not directly for this work.

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Table 1: Subject details for walking speed stratified groups; data is presented as mean ± standard deviation

Group	Number of Subjects	Walking Speed (m/s)	Age (Years)	BMI (kg/m^2)	Years to Review	Gender (M:F)
Slower	11	0.72 ± 0.09	74.18 ± 7.21	29.81 ± 4.99	10.00 ± 0.20	8:3
Faster	11	1.29 ± 0.12	70.56 ± 4.88	25.34 ± 3.51	9.76 ± 0.30	6:5
Control	10	1.36 ± 0.12	64.00 ± 3.60	26.22 ± 3.01	N/A	6:4

Table 2: Kinematics comparison between slower THR, faster THR and controls. Data is presented as mean ± standard deviation. ***=p<0.001, **=p<0.01 and *=p<0.05 for differences between THR groups and control group only.

	Maximum hip extension (°)	Hip Adduction range (°)	Hip rotation range (°)
Slower	-2.8 ± 11.1***	8.0 ± 4.1***	18.6 ± 6.8
Faster	-15.5 ± 7.5	10.9 ± 3.3**	23.0 ± 6.7
Control	-19.4 ± 6.1	13.8 ± 2.3	20.6 ± 6.4

Table 3: Moments comparison between slower THR, faster THR and controls. Data is presented as mean ± standard deviation. ***=p<0.001, **=p<0.01 and *=p<0.05 for differences between THR groups and control group only.

	Max. flexion moment (Nm/kg)	Max. extension moment (Nm/Kg)	HS abduction moment (Nm/kg)	MS abduction moment (Nm/kg)	TO abduction moment (Nm/kg)	Maximum internal rotation moment (Nm/kg)
Slower	0.33 ± 0.11***	-0.63 ± 0.26	-0.64 ± 0.18*	-0.60 ± 0.15**	-0.68 ± 0.16	0.11 ± 0.07**
Faster	0.59 ± 0.10**	-0.82 ± 0.20	-0.76 ± 0.18	-0.50 ± 0.15	-0.74 ± 0.16	0.13 ± 0.07*
Control	0.76 ± 0.18	-0.83 ± 0.25	-0.82 ± 0.18	-0.44 ± 0.14	-0.76 ± 0.16	0.17 ± 0.07

Table 4: Hip contact forces for all groups; data is presented as mean ± standard deviation. "slower" and "faster" groups are an average of the operated and non-operated hips. ***=p<0.001, **=p<0.01 and *=p<0.05 for differences between THR groups and control group only for slower, faster and control groups. No statistical differences were evident between operated and non-operated hips.

Group	Heel-strike force (BW)	Mid-stance force (BW)	Toe-off force (BW)
Slower	2.98 ± 0.67**	2.70 ± 0.63*	3.18 ± 0.71***
Faster	3.43 ± 0.62	2.35 ± 0.57	4.49 ± 0.94
Control	3.79 ± 0.71	2.17 ± 0.54	4.92 ± 0.93
Slower – Operated	3.08 ± 0.62	2.73 ± 0.68	3.24 ± 0.74
Slower – Non-Operated	2.89 ± 0.68	2.67 ± 0.55	3.13 ± 0.68
Faster – Operated	3.22 ± 0.62	2.25 ± 0.58	4.07 ± 0.64
Faster – Non-Operated	3.63 ± 0.53	2.44 ± 0.53	4.91 ± 0.96

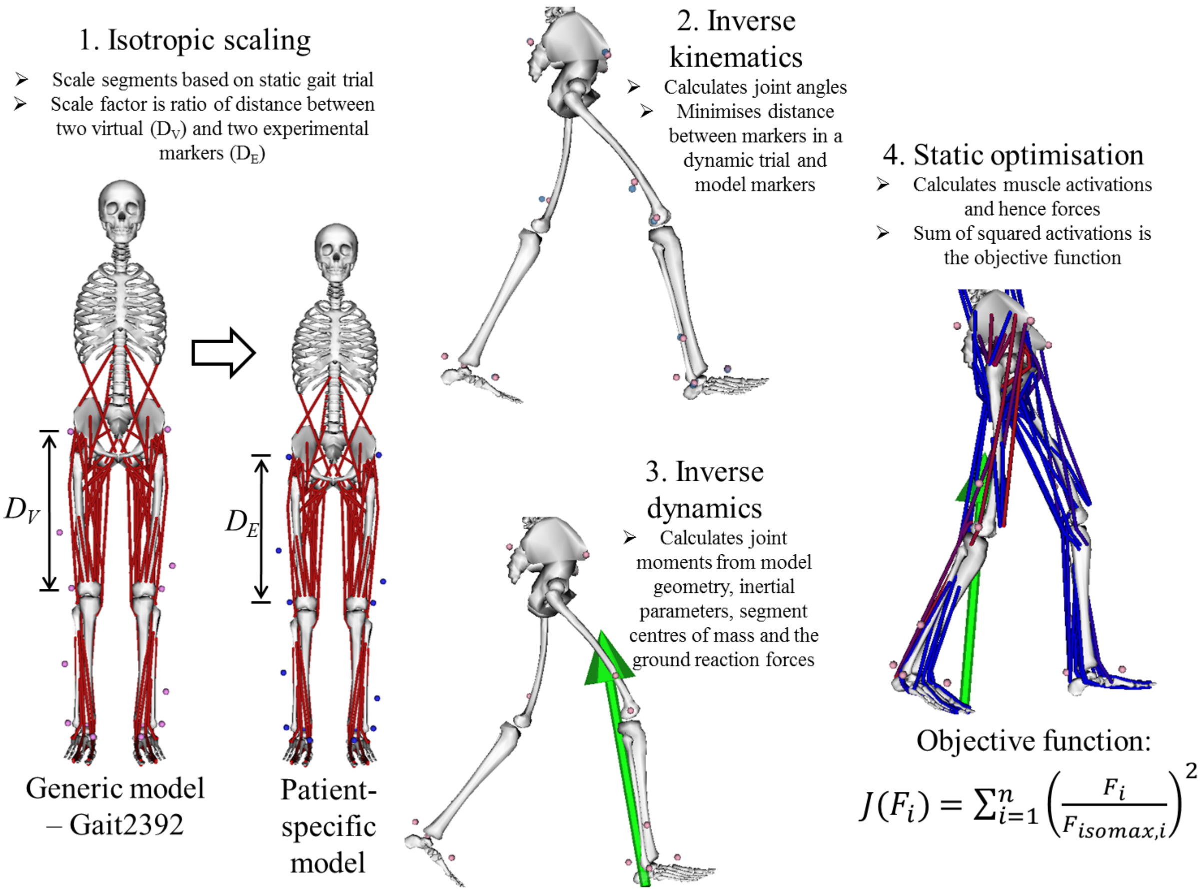

Figure 1: Overview of the workflow for musculoskeletal modelling. In the objective function J, n is the number of muscles in the model, F_i is the force of the i_th muscle and $F_{\mathrm{isomax,i}}$ is the maximum isometric force of the muscle.

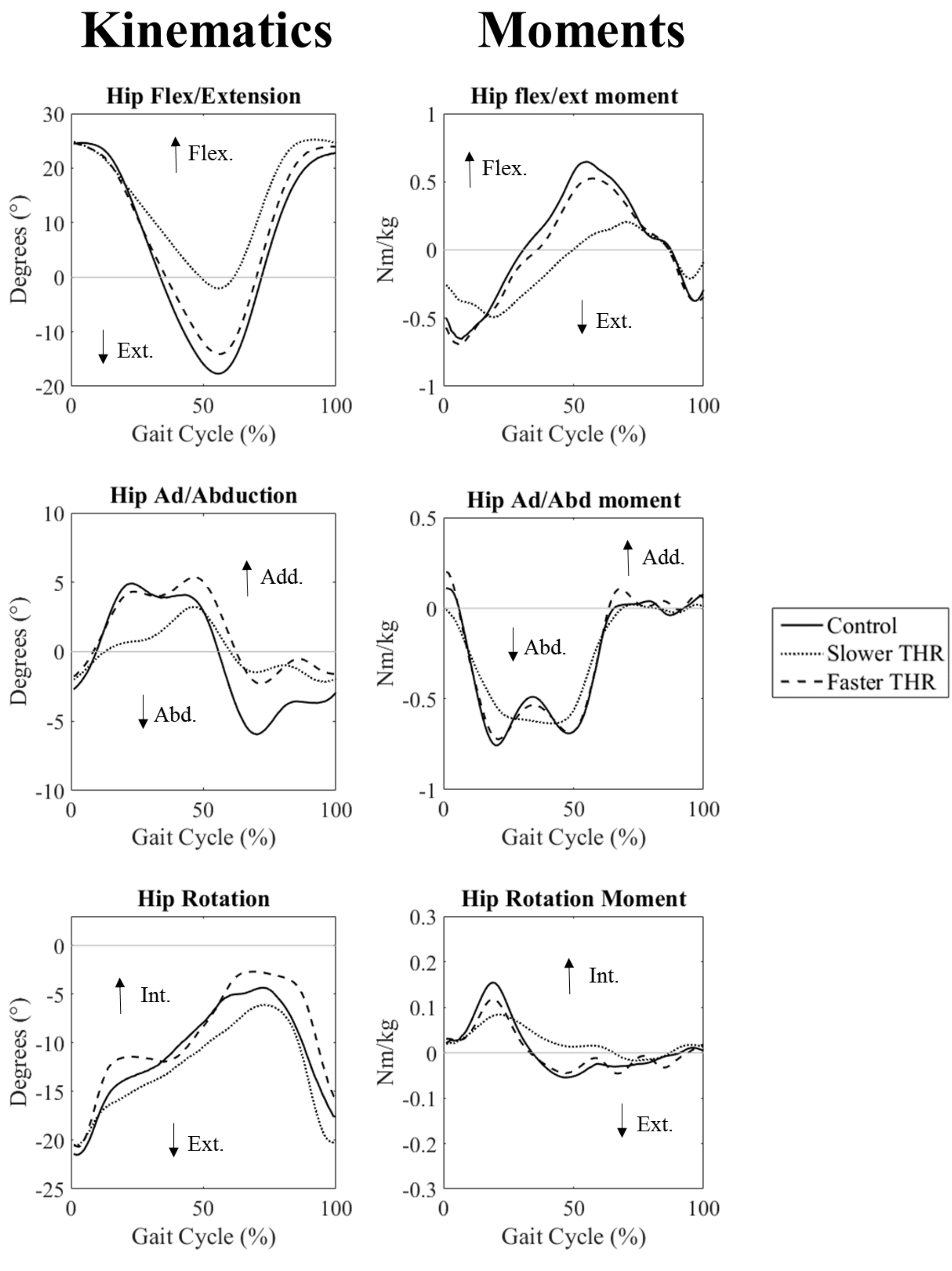

Figure 2: Inverse kinematics and dynamics results for control (solid line), slower THR (dotted), and faster THR (dashed). See Table 2 (kinematics) and Table 3 (moments) for detailed numerical data and statistical testing results.

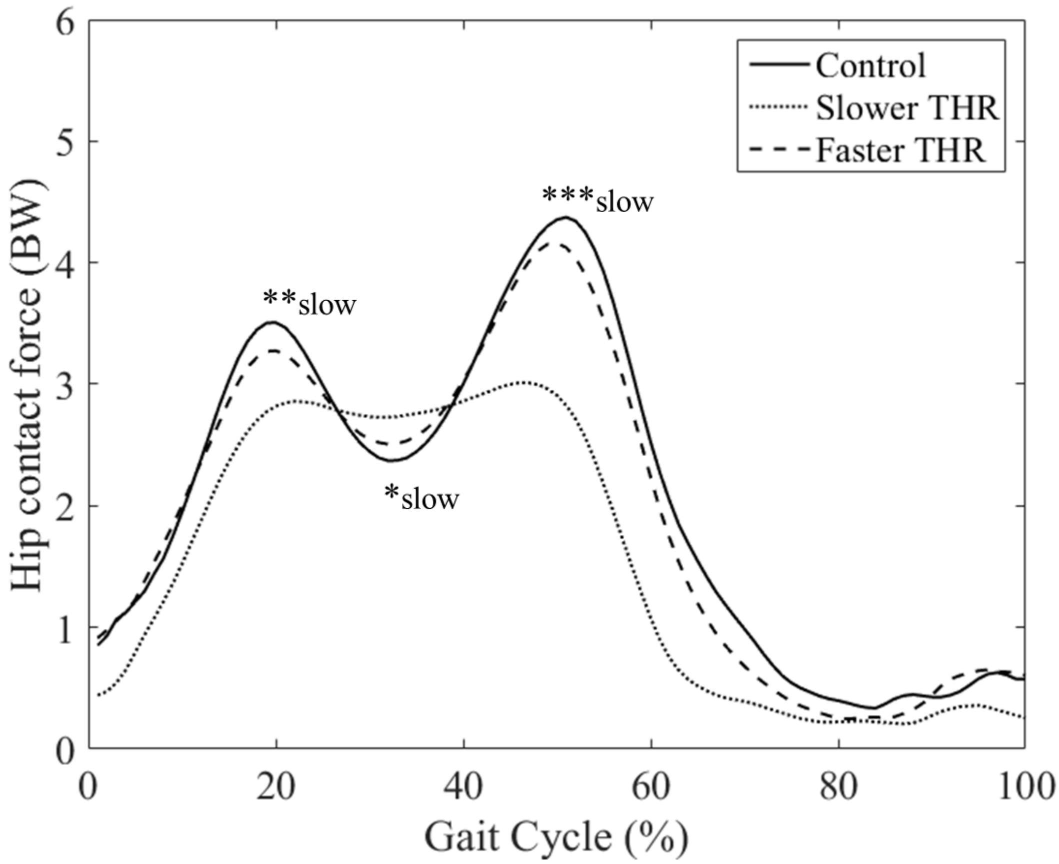

Figure 3: Hip joint contact forces over one gait cycle for slower and faster THR patients compared to elderly control subjects. There were no statistical differences between the control and faster THR groups at any of the three points analysed. Asterisks denote difference (***=p<0.001, **=p<0.01 and *=p<0.05) between controls and speed stratified groups only and are followed by 'slow' for difference between slower and control subjects and 'fast' for differences between faster and control subjects.

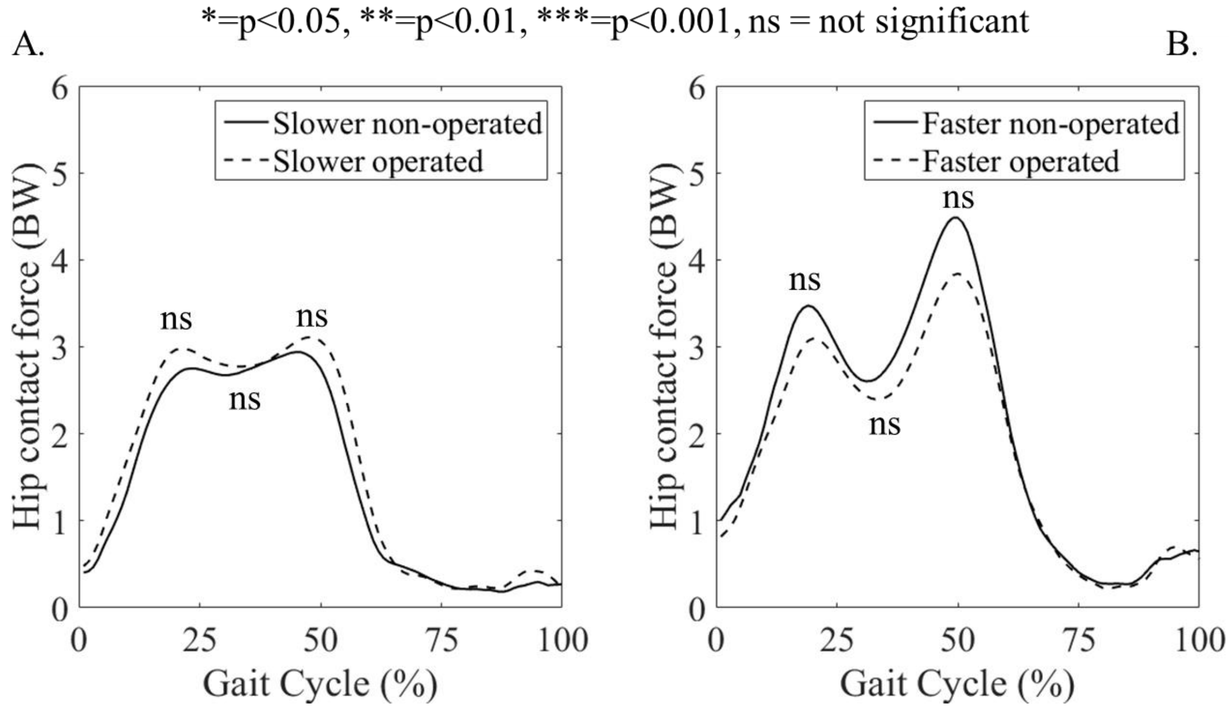

Figure 4: Comparison of operated and non-operated HCFs for slower (A) and faster (B) groups. ‘ns’ refers to differences that were not significant.

Appendix

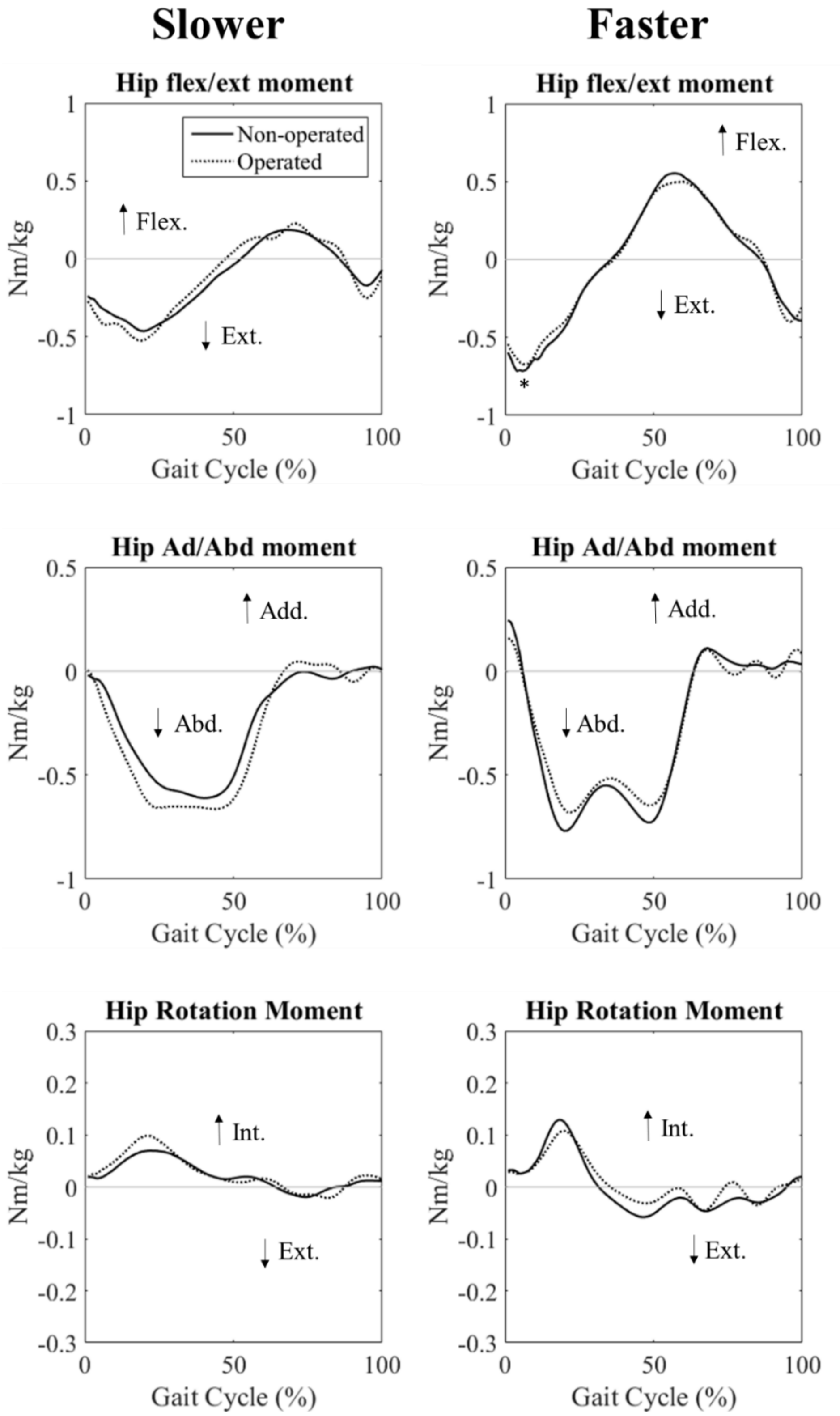

Figure 5: Comparison of moments between operated and non-operated limbs. Maximum hip extension moment in faster THR patients was the only peak with a statistical difference ($p<0.05$) between operated and non-operated hips.