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Musculoskeletal model choice influences hip joint load estimations during gait

Joshua T. Weinhandl [a,*], Hunter J. Bennett [b]

[a] Department of Kinesiology, Recreation, and Sports Studies, The University of Tennessee, United States
[b] Department of Human Movement Sciences, Old Dominion University, United States

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ABSTRACT

The prevalence of musculoskeletal modeling studies investigating hip contact forces and the number of models used to conduct such investigations has increased in recent years. However, the consistency between models remain unknown and differences in model predicted hip contact forces between studies are difficult to distinguish from natural inter-individual differences. The purpose of this study was therefore to evaluate differences in hip joint contact forces during gait between four OpenSim models. These models included the generic models gait2392 and the Arnold Lower Limb Model, as well as the hip specific models hip2372 and London Lower Limb Model. Data from four individuals who have had a total hip replacement with instrumented hip implants performing slow, normal, and fast walking trials were taken from the HIP98 database to evaluate the various models effectiveness at estimating hip loads. Muscle forces were estimated using static optimization and hip contact forces were calculated using the JointReaction analysis in OpenSim. Results indicated that, for gait, the hip specific London Lower Limb Model consistently predicted peak push-off hip joint contact forces with lower magnitude and timing errors compared to the other models. Likewise, root mean square error values were lowest and correlation coefficients were highest for the London Lower Limb Model. These results suggest that the London Lower Limb Model is the most appropriate model for investigations focused on hip joint loading.

1. Introduction

Hip joint loading during activities of daily living such as walking and stair climbing is essential for maintaining healthy bone structure. The relationship between inadequate lower extremity loading and poor bone density, particularly in aging females, is well established (Jamsa et al., 2006; Vainionpaa et al., 2007). However, excessive hip joint loading may increase the likelihood of developing osteoarthritis in healthy hips (Felson, 2013). The direct measurement of internal hip forces developed during human movement, however, is difficult to achieve for practical and ethical reasons. *In vivo* hip contact forces and stresses have been recorded via instrumented prostheses in a limited number of patients (Bergmann et al., 2001, 1993; Damm et al., 2010; Schwachmeyer et al., 2013), and in several cases can be retrieved from the public database www.OrthoLoad.com. Recently, *in silico*, musculoskeletal models have been employed to estimate hip contact forces (Bergmann et al., 2016; Giarmatzis et al., 2015; Heller et al., 2001; Modenese and Phillips, 2012; Shelburne et al., 2010b; Weinhandl et al., 2017). These musculoskeletal models can potentially be implemented without the need for patient medical images. The noninvasive nature of musculoskeletal modeling thereby allows for force estimation with reduced subject risk and financial cost.

In recent years, software packages such as OpenSim (Delp et al., 2007) and Anybody (Damsgaard et al., 2006) have made possible the sharing and distribution of musculoskeletal models. As the number of musculoskeletal modeling based investigations continues to increase, the uniformity in results derived from different models is becoming more critical. Predicted knee contact forces during an idealized knee-extension task have been compared between several, commonly used generic musculoskeletal models (Wagner et al., 2013). They reported that simple scaling and usage of the same objective function for muscle force prediction was not sufficient to produce consistent muscle and knee joint contact forces between models. However, the consistency of various models at predicting hip joint contact forces remains unknown and differences in model predicted hip contact forces between studies are difficult to distinguish from natural inter-individual differences. As

* Corresponding author at: Department of Kinesiology, Recreation, and Sports Studies, The University of Tennessee, 1914 Andy Holt Ave., Knoxville, TN 37996, United States.

E-mail address: jweinhan@utk.edu (J.T. Weinhandl).

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such, with the increased prevalence of musculoskeletal modeling studies investigating hip contact forces it is vital to understand the differences between commonly employed models. While previous models have been validated via instrumented prosthesis (Heller et al., 2001; Modenese et al., 2011), these models have not been compared to one another. The purpose of this study was therefore to evaluate differences in hip joint contact forces during gait between four OpenSim models.

2. Methods

Four musculoskeletal models, freely available in OpenSim (http://opensim.stanford.edu/) were evaluated in this study. These models included two generic gait models and two hip specific models. The generic models were gait2392 (Delp et al., 1990) and the Arnold Lower Limb Model (ALLM) (Arnold et al., 2010). The first hip specific model was hip2372 (Shelburne et al., 2010a, 2010b). This model is based on gait2392 with the addition of the obturator and rectus abdominus muscles. Adjustments were also made to the muscle geometry so that the moment arms of the model matched experimental measurements (Delp et al., 1999; Dostal et al., 1986; Jorgensen et al., 2001; Nemeth and Ohlsen, 1985). Finally, wrapping objects were added for the gluteus maximus, iliacus, and psoas. The second hip specific model was the London Lower Limb Model (LLLM) (Modenese et al., 2011). Muscle attachments and joint kinematics in the LLLM are based on the anatomical dataset published by Klein Horsman et al. (2007). All models were simplified to only include the right leg. Neither contraction dynamics nor force-length-velocity relationships were implemented for the muscle actuators of any model. Other model-defined muscle parameters (e.g., maximum isometric strength, optimal fiber length, pennation angle, etc.) were not changed as the purpose of the current study was to evaluate the default predictive potential of the models. Finally, although all models contain a subtalar joint, the LLLM does not include a metatarsal-phalangeal joint. However, metatarsal-phalangeal and subtalar joints were chosen as "locked" for all models in this study.

Marker coordinates, ground reaction forces, and *in vivo* hip contact forces of four individuals who have had a total hip replacement with an instrumented prosthesis performing slow (n = 11), normal (n = 27), and fast (n = 14) walking trials were taken from the HIP98 database to evaluate the various models effectiveness at estimating hip loads (Table 1). For each patient, segment lengths and muscle attachment sites were scaled using the joint center positions, and segment masses were manually adjusted according to the values reported by Bergmann et al. (2001). Model generalized coordinates that best reproduced the experimental marker coordinate data for each trial were computed by solving an inverse kinematics problem. This global optimization algorithm is formulated as a least-squares problem that minimizes the differences between the measured marker locations and the model's virtual marker locations, subject to joint constraints (Lu and O'Connor, 1999). Inverse dynamics was performed using the measured ground reaction forces and inverse kinematics results to calculate the intersegmental moments. Muscle forces were estimated using static optimization to minimize the sum of squares of muscle activation at each instant in time (Anderson and Pandy, 2001). Idealized torque actuators, known as reserve actuators, were included for each degree of freedom in the model to provide extra torque if the muscles could not generate the measured accelerations. The peak and root-mean-square of each reserve actuator was verified to be less than 5% of the net moment calculated via inverse dynamics (Hicks et al., 2015). Estimated muscle activations were then verified via qualitative comparisons between the model-based predicted activations and experimental electromyographic data provided in the HIP98 dataset as per current recommendations (Hicks et al., 2015). Due to concern that the magnetic field of the coil used for powering the prosthesis could affect the electromyographic signal in the HIP98 dataset, we also compared our model-based activations to experimental electromyographic data available in the literature (Kadaba et al., 1989; Liu et al., 2008). Finally, hip contact forces of the right leg were calculated using the JointReaction analysis in OpenSim (Steele et al., 2012) and reported in the pelvic reference frame (Lewis and Garibay, 2015). Resultant hip contact forces were utilized for model comparisons. This process of scaling, inverse kinematics, inverse dynamics, static optimization, and joint reaction analysis was repeated for all four models.

To assess the similarity between *in vivo* and simulated hip contact forces, root mean square error (RMSE) and Pearson's product-moment correlation coefficients (R) were calculated for the entire gait cycle for each trial and each model. First and second peak hip contact forces, respectively defined as peak loading-response and peak push-off, were also identified for each trial and each model. In the event that two distinct peaks were not evident for a trial, a single peak was identified and assigned as loading-response if it occurred in the first half of stance or push-off if it occurred in the latter half of stance. This occurred in 10 normal walking, 4 fast walking, and 5 slow walking trials, with all instances occurring in the first half of stance and subsequently being assigned as loading-response peaks. Relative deviation was then calculated as the difference between *in vivo* and simulated peak hip contact forces divided by the *in vivo* hip contact force. Time shift (Fig. 1) between the *in vivo* and simulated peak hip contact forces was also calculated and expressed as a percentage of gait cycle.

3. Results

Ensemble hip joint contact forces for each model and the HIP98 dataset for slow, normal, and fast walking are presented in Fig. 2. Individual patient ensemble hip joint contact forces for each model are presented in Fig. 3. Pearson's correlation coefficients indicate a strong correlation between experimentally measured and simulated hip contact forces for all models regardless of speed (Table 2). During slow walking trials LLLM yielded a higher correlation coefficient than ALLM, hip2372, and gait2392. Likewise, gait2392 yielded a higher correlation coefficient than ALLM and hip2372.

Table 1
Characteristics of the four patients taken from HIP98 (Bergmann et al., 2001).

					Walking speeds (m/s)		
Subject	Sex	Age	Weight (N)	Height (m)	Slow	Normal	Fast
HSR	M	55	860	1.74	1.04 (n = 1)	1.36 (n = 8)	1.64 (n = 5)
KWR	M	61	702	1.65	1.05 (n = 5)	1.15 (n = 8)	1.40 (n = 5)
PFL	M	51	980	1.75	1.02 (n = 5)	1.13 (n = 6)	1.40 (n = 4)
IBL	F	76	800	1.70	–	1.08 (n = 5)	–

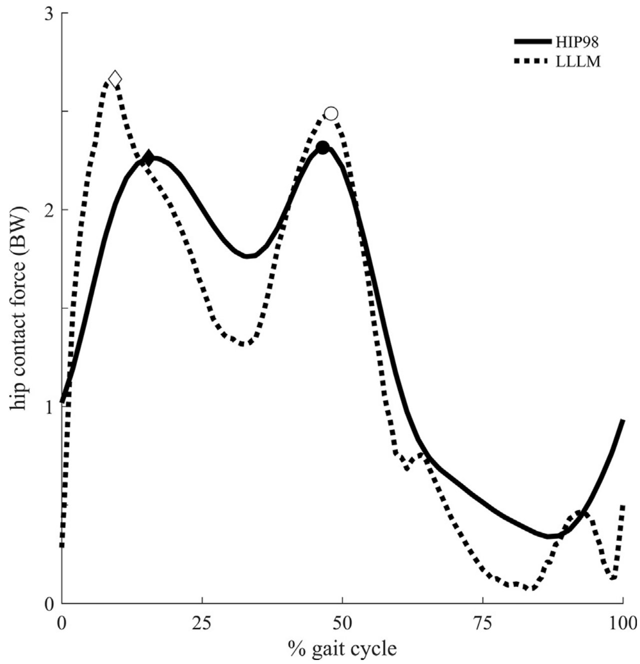

Fig. 1. Experimental (solid line) and simulated (dashed line) hip contact forces during a representative normal walking trial for patient KWR. Diamond markers indicate loading-response peaks and circle markers represent push-off peaks. Relative deviation was calculated as the difference between experimental and simulated peak hip contact forces divided by the experimental hip contact force. Time shift between the experimental and simulated hip contact force peaks was calculated and expressed as a percentage of gait cycle.

A similar pattern was observed for normal walking trials. However, all models exhibited similar correlation coefficients for fast walking trials.

For the LLLM model, RMSE was lower during the slow and normal walking trials than during the fast walking trials. The RMSE for LLLM during slow and normal was also lower than the other models. Interestingly, the RMSE was lowest during normal walking trials for gait2392, hip2372 and ALLM compared to slow and fast walking trials. However, they were still considerably larger than LLLM. For fast walking trials all models displayed similar RMSE values.

With regards to discrete measures, all models exhibit similar ability to predict loading-response peak hip contact force magnitude. This magnitude error consistently decreased as walking speed increased for all models. Likewise, there was no difference in timing error for the loading-response peak between the models, with all model predictions occurring early by 2.2–4.1% of the gait cycle. There were, however, differences between models for peak push-off hip contact force magnitude and timing errors.

During slow walking trials, error in LLLM model predictions of peak push-off hip contact force magnitude was considerably lower than all other models. All models predictions of peak push-off hip contact force were earlier than those measured experimentally. However, LLLM timing error ($-1.5 \pm 1.7\%$) was lower than gait2392 (-3.1 ± 1.5), hip2372 (-3.2 ± 1.2), and ALLM (-3.5 ± 1.4).

LLLM once again predicted peak push-off hip contact force with the smallest magnitude errors compared to all other models during normal walking trials. With regard to timing error, LLLM predicted peak push-off hip contact force 0.6% of the gait cycle after the experimental peak while all other models predicted it 1.1–1.5% before the experimental peak.

Finally, peak push-off hip contact force magnitude error was lowest for the LLLM model compared to the other models during

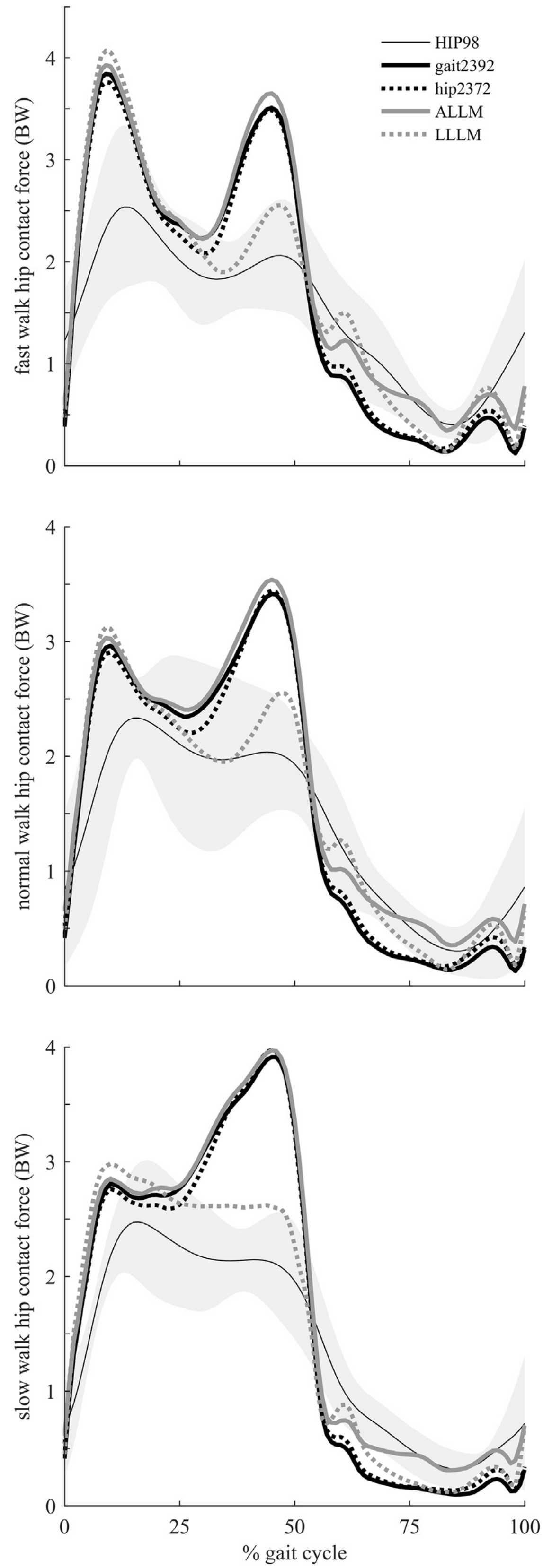

Fig. 2. Ensemble mean hip contact forces for slow, normal, and fast walking trials. Experimental data includes four total hip replacement patients from the HIP98 database (Bergmann et al., 2001). Models evaluated include gait2392 (solid black lines), hip2372 (dashed black lines), Arnold Lower Limb Model (ALLM, solid grey lines), and London Lower Limb Model (LLLM, dashed grey lines).

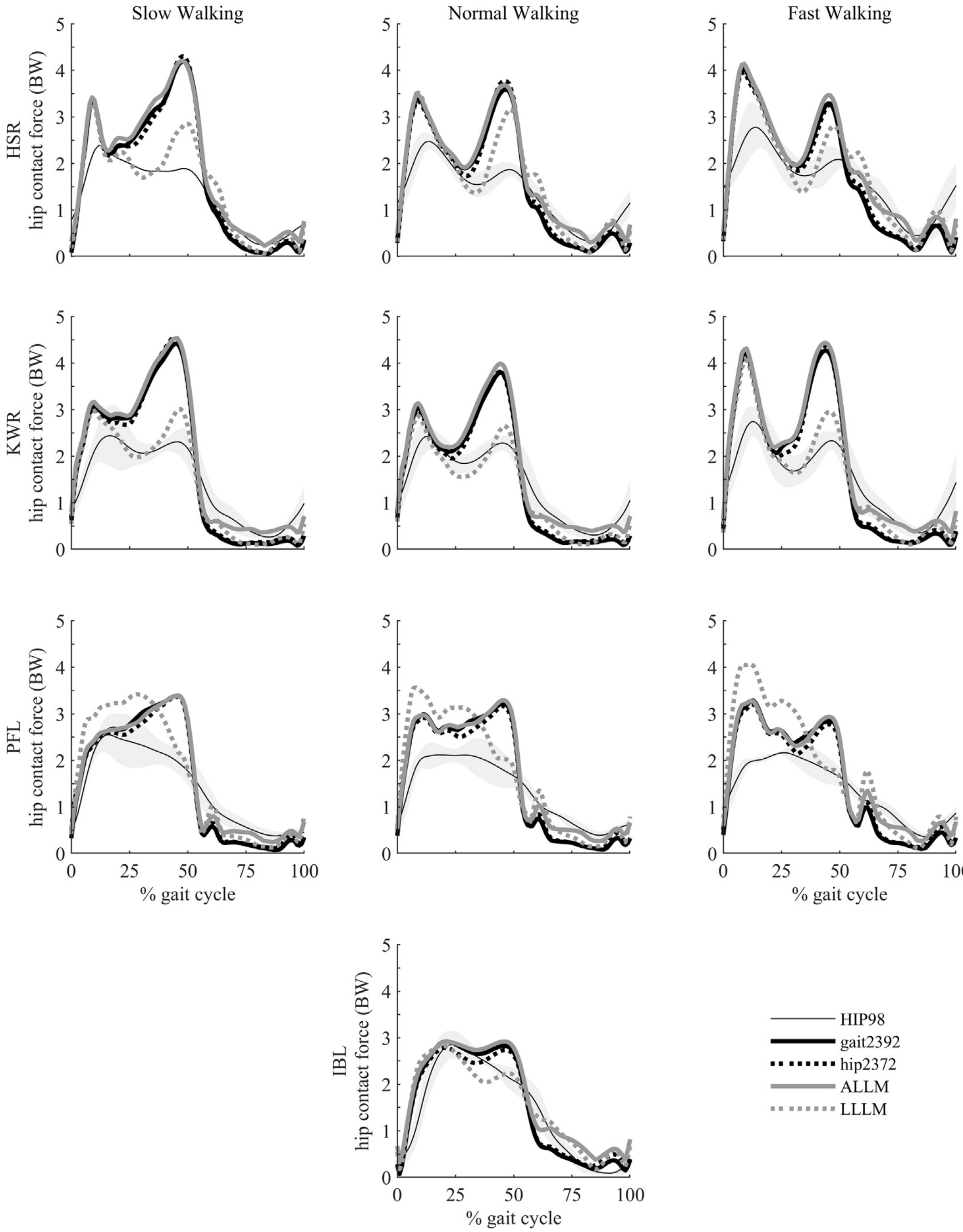

Fig. 3. Subject ensemble mean hip contact forces for slow, normal, and fast walking trials. Experimental data includes four total hip replacement patients from the HIP98 database (Bergmann et al., 2001). Models evaluated include gait2392 (solid black lines), hip2372 (dashed black lines), Arnold Lower Limb Model (ALLM, solid grey lines), and London Lower Limb Model (LLLM, dashed grey lines).

fast walking trials. Interestingly, peak push-off hip contact force magnitude error remained relatively constant across speeds for LLLM while increasing as speed increased for the other models. With respect to timing error, LLLM predicted the peak push-off hip contact force 1.4% of the gait cycle after the experimental peak, while all other models predicted it 1.4–1.5% of the gait cycle before the experimental peak.

4. Discussion

The objective of this study was to evaluate differences in hip joint contact force during gait between four OpenSim models: gait2392 (Delp et al., 1990), ALLM (Arnold et al., 2010), hip2372 (Shelburne et al., 2010a, 2010b), and LLLM (Modenese et al., 2011). To accomplish this objective, gait of four individuals who have had a total hip replacement with an instrumented prosthesis (Bergmann et al., 2001) was simulated using each model. The simulated hip joint forces were then compared to the experimental hip joint forces, with LLLM predicting peak push-off hip contact forces with lower magnitude and timing errors compared to the other models evaluated.

The RMSE values generally increased while correlation coefficients generally decreased as walking speed increased for LLLM model. This finding is consistent with Modenese and Phillips

Table 2
Pearson's correlations coefficient (R), root mean square error (RMSE), peak loading-response hip contact force magnitude and timing errors, and peak push-off hip contact force magnitude and timing errors for each model and walking speed. Data are presented as mean ± standard deviation.

		gait2392	hip2372	ALLM	LLLM
R	Slow	0.93 ± 0.02	0.91 ± 0.02	0.92 ± 0.02	0.95 ± 0.02
	Normal	0.92 ± 0.03	0.91 ± 0.03	0.91 ± 0.04	0.93 ± 0.02
	Fast	0.90 ± 0.02	0.90 ± 0.02	0.89 ± 0.02	0.90 ± 0.03
RMSE (%BW)	Slow	82.2 ± 19.4	81.6 ± 22.8	82.2 ± 23.7	50.0 ± 13.2
	Normal	66.7 ± 14.8	65.0 ± 15.9	68.6 ± 13.7	53.4 ± 11.5
	Fast	81.3 ± 14.1	77.2 ± 16.1	80.2 ± 16.5	79.3 ± 7.7
Loading-response	Slow	46.2 ± 13.6	42.9 ± 13.6	49.1 ± 13.5	41.5 ± 24.3
Peak magnitude error	Normal	27.7 ± 16.7	24.7 ± 17.1	29.6 ± 17.4	25.1 ± 20.1
(% experimental peak)	Fast	20.5 ± 16.0	18.1 ± 16.1	22.3 ± 17.4	21.7 ± 17.0
Loading-response	Slow	−2.2 ± 3.6	−2.3 ± 3.3	−2.3 ± 3.0	−2.5 ± 3.1
Peak timing error	Normal	−3.3 ± 2.8	−3.5 ± 2.7	−3.2 ± 3.0	−4.1 ± 2.1
(% gait cycle)	Fast	−3.5 ± 3.7	−3.5 ± 3.6	−3.6 ± 3.8	−4.1 ± 3.6
Push-off	Slow	75.1 ± 18.9	77.4 ± 21.0	82.8 ± 17.7	33.8 ± 9.7
Peak magnitude error	Normal	83.0 ± 20.6	88.1 ± 25.0	89.3 ± 18.1	45.0 ± 13.5
(% experimental peak)	Fast	98.3 ± 15.3	105.7 ± 15.3	102.3 ± 13.6	35.4 ± 13.2
Push-off	Slow	−3.1 ± 1.5	−3.2 ± 1.2	−3.5 ± 1.4	−1.5 ± 1.7
Peak timing error	Normal	−1.1 ± 1.7	−1.5 ± 1.5	−1.4 ± 1.6	0.6 ± 1.9
(% gait cycle)	Fast	−1.4 ± 1.3	−1.5 ± 1.2	−1.4 ± 1.3	1.4 ± 2.0

(2012), suggesting that the global goodness of fit decreases as gait speed increases for the LLLM. This was not the case for the gait2392, hip2372, or ALLM models, which showed the lowest RMSE during normal walking. For these models, RMSE was similar for slow and fast walking trials. This is most likely due to the exaggerated loading-response and push-off hip joint force peaks exhibited by these models, particularly the push-off hip joint force peak during slow walking.

With regard to peak hip contact force magnitudes and timings, the LLLM generally resulted in lower errors. However, the lowest error for loading-response during normal and fast walking was observed with the hip2372 model. All models exhibited loading-response peak magnitude errors of 41.5–49.1% of the experimental peak during slow trials, 24.7–29.6% during normal trials, and 18.1–22.3% for fast trials. Modenese et al. (2011) reported an average difference of 20.8% between experimental and simulated peaks during normal walking, which is consistent with the current study's findings. On the other hand, Modenese and Phillips (2012) reported much lower errors of approximately 8%, 9% and 15% for slow, normal and fast walking trials, respectively. The apparent discrepancy between studies is most likely due to how error values were calculated. Modenese and Phillips (2012) reported differences in simulated hip contact forces at the instant of the experimental peak, whereas Modenese et al. (2011) and the current study reported peak-to-peak deviations between experimental and simulated hip contact forces. The former does not account for timing errors between the simulated and experimental data. While timing errors in the current study were small, 2–4% for loading-response peak and 1–3% for push-off peak, these differences explain the differences in error values reported by Modenese and Phillips (2012) and the current study. Furthermore, Modenese et al. (2011) and Modenese and Phillips (2012) reported a singular, global maximum hip contact force, while two local maximums were identified in the current study. The global maximum most likely corresponds to the loading-response peak of the current study as the experimental push-off peak magnitude was generally less than the loading-response peak. This feature was accurately predicted by only the LLLM model, as all other models exhibited higher push-off peaks during normal and slow walking trials compared to the loading-response peaks.

Interestingly, the largest differences between models were observed during the second half of stance. At the loading-response peak, all models had similar magnitude error scores. However, peak push-off magnitude errors for the gait2392, hip2372 and ALLM models were 2–3 times higher than the peak push-off magnitude errors for the LLLM. This is potentially due to differences in muscle moment arms between the models, particularly when the hip is in an extended position as it is at the push-off peak. Studies have shown that accurate representation of a muscle path improves model estimates of moment arms. Inaccurate moment arm estimates would lead to inaccurate muscle force estimates to reproduce the intersegmental moments. For example, underestimating moment arms would like to an overestimation of muscle forces, thereby increasing the hip contact force (Modenese et al., 2013).

Evaluating the moment arm estimates of the models used in this study it is evident that there are substantial differences that most assuredly influence hip contact force estimates (Figs. 4 and 5). For most muscles, there were minimal differences in moment arms between gait2392, hip2372, and ALLM. This was expected as the muscle paths for these models are all based on the original muscle paths provided by Delp et al. (1990), which was based on the anatomical dataset published by Brand et al. (1982). The primary modifications in hip2372 were adjustments to the muscle geometry and inclusion of wrapping objects for the gluteus maximus so that the moment arms of the model matched the measurements of Delp et al. (1999), Nemeth and Ohlsen (1985), and Dostal et al. (1986). Whereas, modifications in ALLM were made based on the measurements of 21 cadaver subjects by Ward et al. (2009). On the other hand, muscle attachments in the LLLM model are based on the anatomical dataset published by Klein Horsman et al. (2007). The most noticeable difference is that there are 163 actuators included in LLLM in order to represent 38 muscles, divided into 57 muscle parts composed of up to 6 bundles. For example, the LLLM model represents the gluteus maximus via 12 bundles compared to the 3 bundle representation of the other models. Muscle paths, and geometric wrapping surfaces which are defined from magnetic resonance imaging or other high-resolution medical images may produce model moment arms that are more closely matched to moment arms measure experimentally (De Pieri et al., 2018; Wesseling et al., 2016). Wesseling et al. (2016) demonstrated that inclusion of patient-specific wrapping surfaces has the largest effect on hip contact forces. Thus, while not patient-specific, the muscle geometry and wrapping surfaces in the LLLM model may explain the improved accuracy during late stance compared to the other models evaluated.

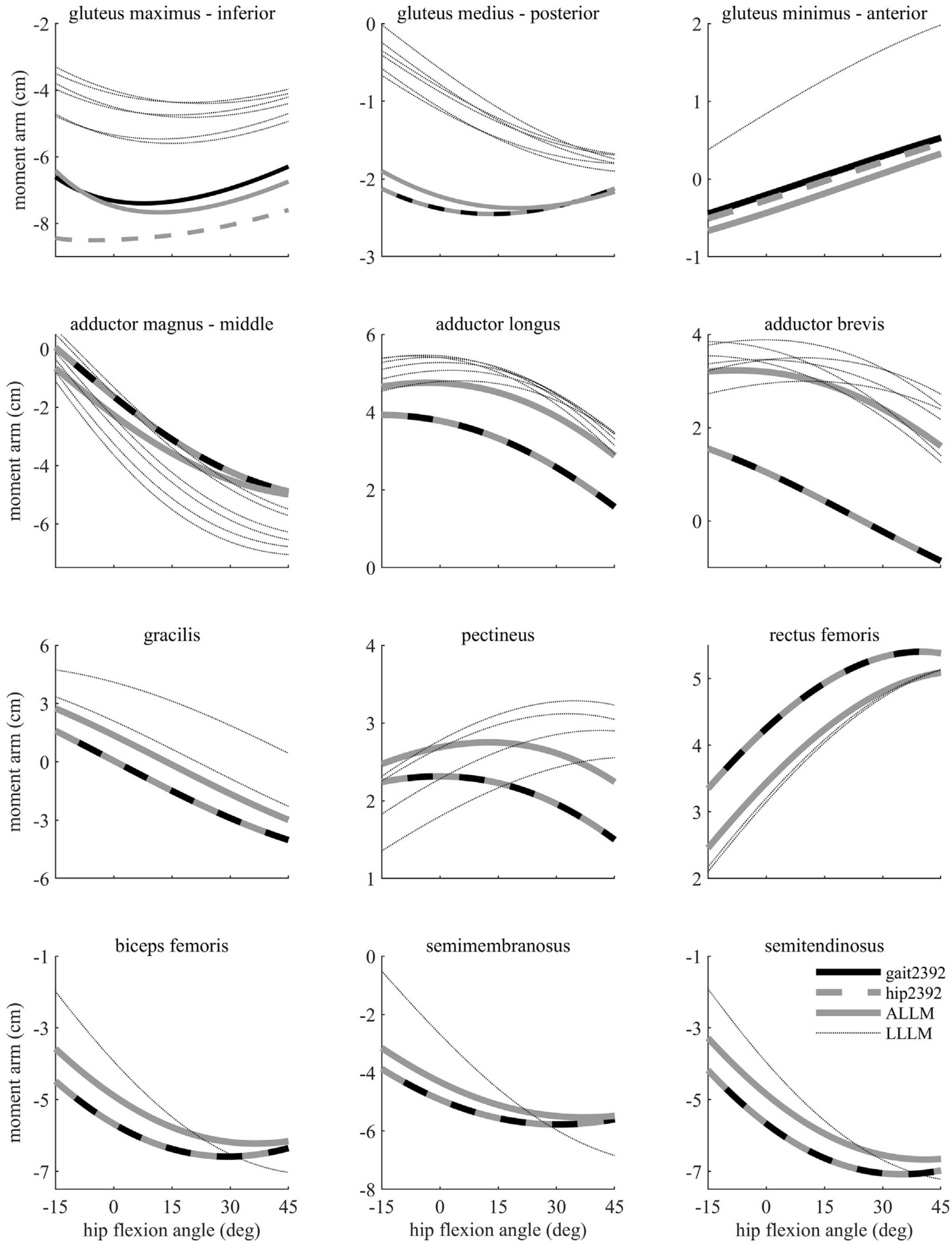

Fig. 4. Sagittal plane moment arms for selected muscles. Models evaluated include gait2392 (solid black lines), hip2372 (dashed grey lines), Arnold Lower Limb Model (ALLM, solid grey lines), and London Lower Limb Model (LLLM, thin black lines). Positive moment arms indicate hip flexors and negative moment arms indicate hip extensors.

Skeletal segment parameters are also important to consider when comparing simulation models. Although several differences in femoral parameters exist between the four models in the current study, one component, hip joint center locations, may be an important factor in contact force predictions. For a 1.75 m tall individual the gait2392 and hip2372 place the hip joint center 7.07 cm posterior, 6.61 cm distal, and 8.35 cm lateral to the midpoint of the anterior superior iliac spines. On the other hand, the ALLM model places the hip joint center 5.63 cm posterior, 7.85 cm distal, and 7.73 cm lateral, and the LLLM hip joint center is 3.76 cm posterior, 8.78 cm distal, and 8.35 cm lateral to the anterior superior iliac spines midpoint. Although alterations to joint center locations have been shown to impact inverse dynamics (Bennett et al., 2018; Kirkwood et al., 1999; Stagni et al., 2000), there were minimal differences in the hip joint moments between the models (Fig. 6). Consequently, the differences in hip joint center locations between models appears to influence moment arm estimation more than intersegmental moment calculations. Future research focusing on hip joint contact force predictions should consider the impact of muscle geometry and skeletal/bony structure parameters.

There are several limitations that must be considered when interpreting the results of the current study. First, although many model parameters such as segment masses and lengths were

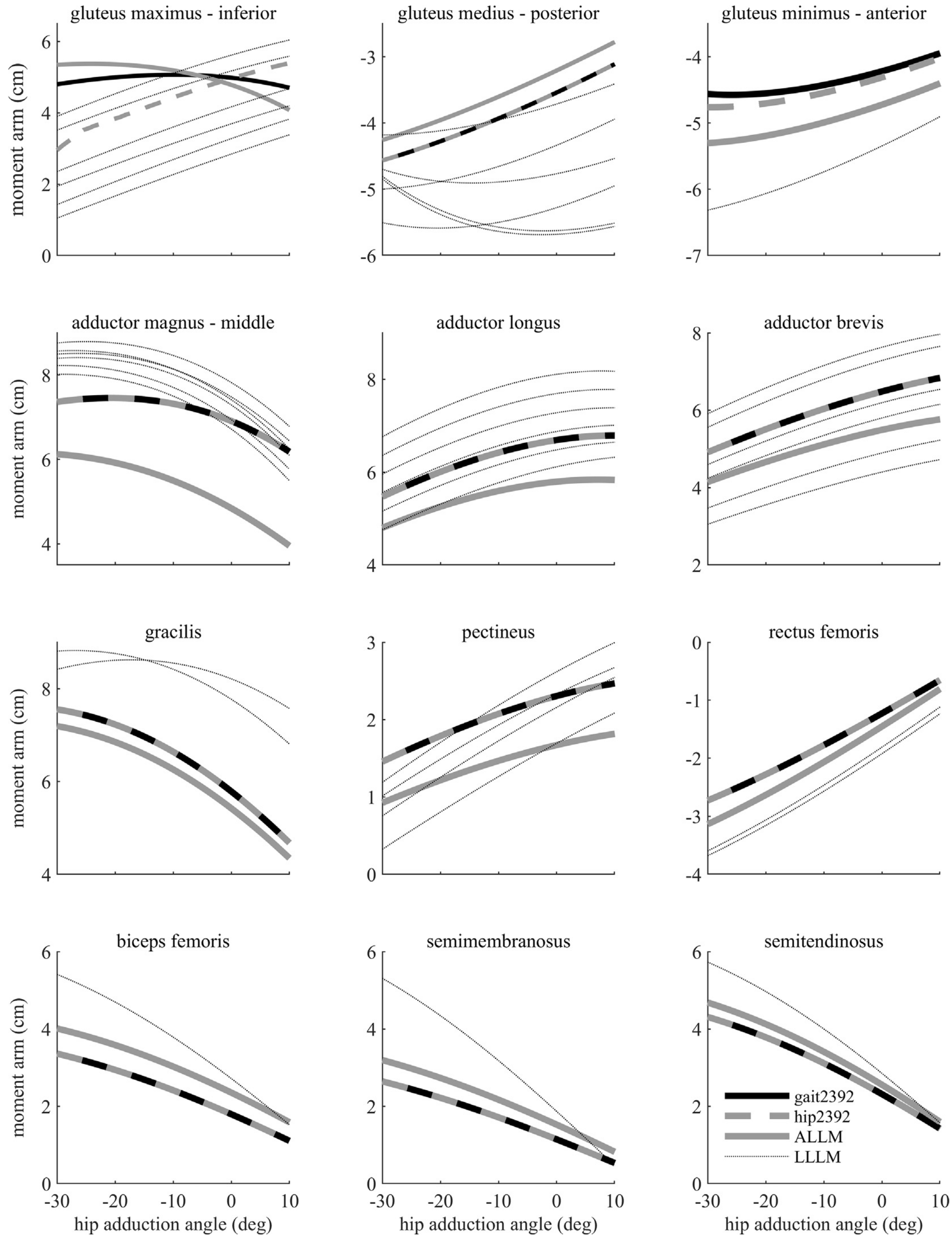

Fig. 5. Frontal plane moment arms for selected muscles. Models evaluated include gait2392 (solid black lines), hip2372 (dashed grey lines), Arnold Lower Limb Model (ALLM, solid grey lines), and London Lower Limb Model (LLLM, thin black lines). Positive moment arms indicate hip adductors and negative moment arms indicate hip abductors.

patient-specific estimates, several parameters defining muscle function such as maximum isometric force, tendon slack length, and optimal fiber length were not subject-specific, but rely on published values measured on a limited sample of cadaveric specimens. Furthermore, neither contraction dynamics nor force-length-velocity relationships were implemented for muscle actuators. The intended use of the LLLM is for estimating muscle forces via static optimization without including the force-length-velocity relationships (Modenese et al., 2015). Therefore, we chose not to include either contraction dynamics or force-length-velocity relationships in any of the models. Anderson and Pandy (2001) demonstrated that the influence of muscle dynamics on muscle force estimation is negligible during normal walking, but this may not be the case for fast walking. Furthermore, while other optimization techniques such as Computed Muscle Control are available, Wesseling et al. (2015) demonstrated that static optimization techniques produced results closest to experimentally measured hip contact forces. Second, skeletal geometry such as femoral anteversion and neck length have strong influences on predicted hip joint forces (Lenaerts et al., 2008). Skeletal morphology, including anteversion and shaft angles, are provided in the HIP98 dataset. Thus, future work could incorporate patient specific skeletal parameters. Finally, static optimization with an objective function that minimized the sum of muscle activations squared was used to estimate

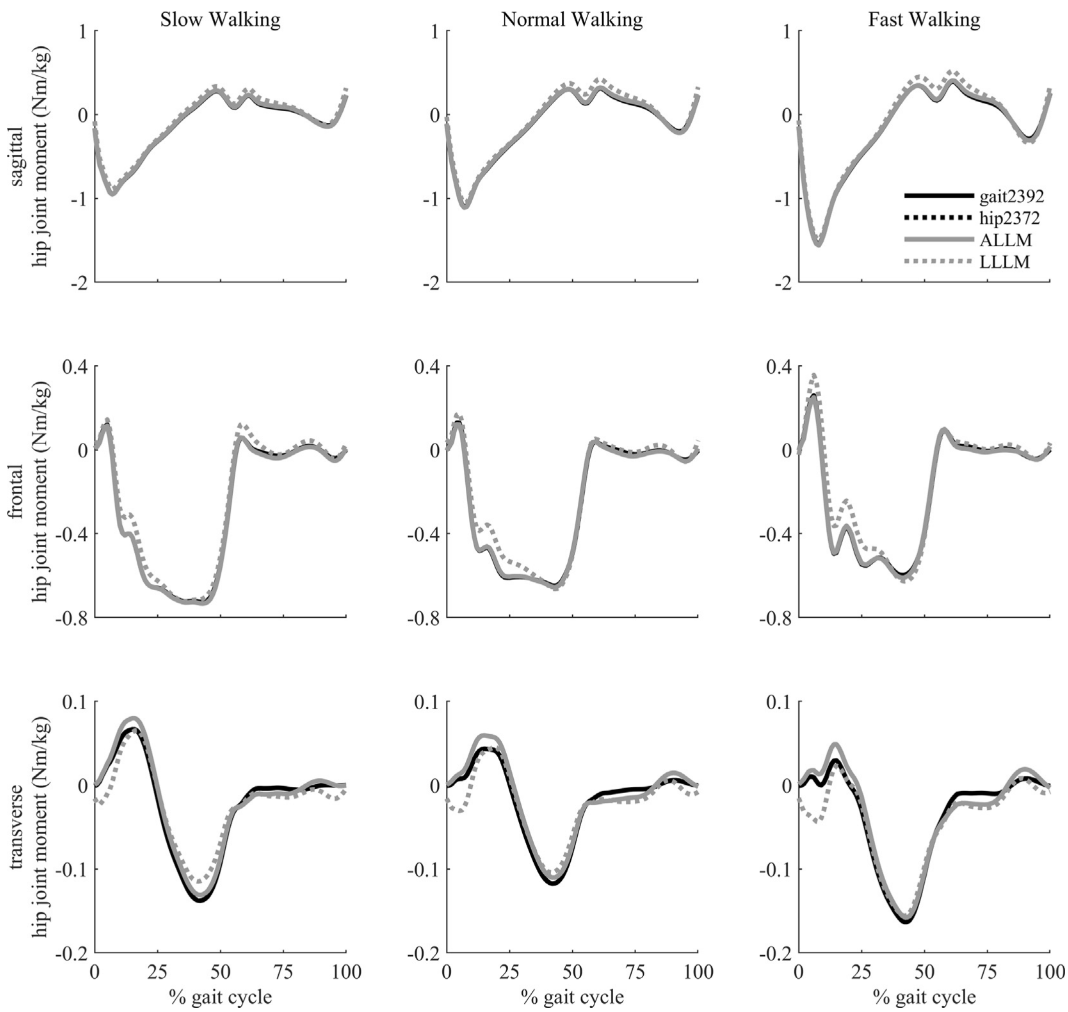

Fig. 6. Ensemble mean hip joint moments for slow, normal, and fast walking trials. Experimental data includes four total hip replacement patients from the HIP98 database (Bergmann et al., 2001). Models evaluated include gait2392 (solid black lines), hip2372 (dashed black lines), Arnold Lower Limb Model (ALLM, solid grey lines), and London Lower Limb Model (LLLM, dashed grey lines). Positive joint moments indicate hip flexion, adduction, and internal rotation.

muscle forces. Static optimization with this objective function has been shown to accurately reproduce muscle activation patterns during walking (Anderson and Pandy, 2001), but hip joint force estimates may have been improved with other objective functions. Lerner et al. (2014) and Steele et al. (2012) reported minimal errors between experimental and simulated knee joint contact forces using an objective function that minimized the sum of squared muscle activations while incorporating individual muscle weighting values. Others have included joint contact forces in the objective function. Demers et al. (2014) demonstrated that minimizing the sum of squared muscle activations produced knee joint contact forces greater than those measured experimentally, while an objective function that minimized vertical knee joint contact force yielded lower model-predicted forces than those measured experimentally. On the other hand, an objective function which minimized both muscle activations and joint loads more accurately recreated the experimental knee joint contact forces than minimizing muscle activations alone (Demers et al., 2014). It is reasonable to assume that errors in the current study could be reduced for all models if hip joint loads were included in the objective function. We chose to use the objective function which only minimizes the sum of squared muscle activations as this is the default objective function for static optimization provided by OpenSim.

In summary, we found the hip specific LLLM consistently predicted peak push-off contact forces with lower magnitude and timing errors compared to gait2392, ALLM, and hip2372. Likewise, RMSE values were lowest and correlation coefficients were highest for LLLM. However, the estimates of hip contact forces for all models in the current study may have been improved by incorporating individual muscle weighting values or joint contact forces in the objective function. These results suggest that, of the models tested, the LLLM most closely reproduces the hip contact forces measured *in vivo* and is the most appropriate model for investigations focused on hip joint loading during gait.

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Declaration of Competing Interest

The authors have no conflict of interest related to the present work to disclose.

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