Gait Alterations to Effectively Reduce Hip Contact Forces

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ABSTRACT: Patients with hip pathology present alterations in gait which have an effect on joint moments and loading. In knee osteoarthritic patients, the relation between medial knee contact forces and the knee adduction moment are currently being exploited to define gait retraining strategies to effectively reduce pain and disease progression. However, the relation between hip contact forces and joint moments has not been clearly established. Therefore, this study aims to investigate the effect of changes in hip and pelvis kinematics during gait on internal hip moments and contact forces which is calculated using muscle driven simulations. The results showed that frontal plane kinetics have the largest effect on hip contact forces. Given the high correlation between the change in hip adduction moment and contact force at initial stance ($R^2 = 0.87$), this parameter can be used to alter kinematics and predict changes in contact force. At terminal stance the hip adduction and flexion moment can be used to predict changes in contact force ($R^2 = 0.76$). Therefore, gait training that focuses on decreasing hip adduction moments, a wide base gait pattern, has the largest potential to reduce hip contact forces. © 2015 Orthopaedic Research Society. Published by Wiley Periodicals, Inc. J Orthop Res 33:1094–1102, 2015.

Keywords: hip contact force; hip moments; gait retraining

Patients with osteoarthritis (OA) present gait alterations[1–3] that often persist after surgical intervention, i.e., total joint replacement.[1,4,5] The gait pattern adopted by these patients influences joint moments and loading. The relation between joint loading and kinematics has been established in patients with knee OA and offers the potential to change the patients' gait pattern as part of a physical therapy treatment to deliberately reduce knee contact forces.[6,7] As clinical studies often assume a relation between the external knee adduction moment and medial knee loading,[7,8] gait retraining strategies were developed that decrease the knee adduction moment and therefore knee loading.[9,10] During training, direct feedback on knee joint loading can be provided by monitoring the external knee adduction moment.[11,12] The success of these gait retraining strategies on pain and disease progression in patients with knee OA is currently being explored.[6,13]

However, for the hip joint, the relation between hip and pelvis kinematics and hip joint loading has not yet been clearly defined. This is relevant as patients with hip impairment can present gait aberrations, such as reduced maximal hip extension and decreased range of hip ab-/adduction[1,2] that can persist up to 10 years after surgery[4]. Also, changes in hip rotation have been reported as well as decreased pelvic stability[2] inducing Trendelenburg gait, i.e., excessive contralateral pelvis drop in combination with increased hip adduction.[14] Furthermore, decreased sagittal plane range of motion and decreased hip adduction have been associated with reduced hip flexion/extension[15] and hip adduction moments.[1]

Some of these clinically observed changes in hip and pelvis kinematics were related to changes in hip joint loading[2,16] Lenaerts et al.[2] linked the combination of a reduced hip adduction and increased external rotation as well as an excessive pelvic obliquity to a decreased hip contact force 6 weeks after total hip arthroplasty (THA). Likewise, Foucher et al.[16] reported that both peak hip adduction and flexion moment were related to hip contact forces in patients after THA, where increased hip contact forces were found with increasing moments. Also, increased hip extension has been shown to increase the anterior hip contact force.[17,18]

THA patients showed decreased hip contact forces[19] compared to control subjects. These differences were attributed to changes in gait speed, cadence, and stride length; however, no link was made to change in kinematics. Previous research also showed that using crutches decreases the magnitude of the hip contact force.[20] Furthermore, Valente et al.[21] found that with increasing abductor weakness a normal gait pattern could be maintained, however, resulting in an increase in hip contact forces at the first peak and a decrease at the second peak.

Besides its magnitude, the orientation of the hip contact force vector is also affected by changes in kinematics. Lenaerts et al.[2] reported a more vertical orientation of the contact forces in patients after THA, indicating a return toward normal loading. Furthermore, these changes in HCF and orientation affect the stress and femoral density distribution[22,23] and therefore, the stability and wear of a prosthesis.[24,25] However, the contribution of the specific pelvis and hip kinematics to the magnitude and direction of the hip contact forces during gait has not been investigated.

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In this study, we investigate to what extent impaired hip and pelvis kinematics can affect the magnitude and orientation of hip contact forces using muscle driven simulations of gait. To do so, we systematically imposed perturbations in hip and pelvis kinematics, including typical gait deviations previously reported in patients. Using this synthetic dataset, we calculate the effect of these gait patterns on the magnitude and orientation of the hip contact forces. The changes that result in clear changes in hip contact forces can then be used to define gait retraining strategies to effectively influence hip loading[2,16] and, therefore, influence disease progression or implant survivorship. Furthermore, we investigated whether hip contact forces can be estimated based upon external joint moments that are readily available in most state of the art clinical gait laboratories. If confirmed, the specific external joint moment could then be used as a representative online feedback signal during the gait retraining session.

METHODS

Experimental Procedure

Two male and three female healthy control subjects (age 56 ± 3 years., range 52–61 years.; BMI $22.3 \pm 1.59\,\mathrm{kg/m^2}$, range $20.6–24.0\,\mathrm{kg/m^2}$) were included in the study and signed an informed consent. The study was approved by the local ethics committee. All subjects walked at self-selected speed (walking speed $1.28 \pm 0.13\,\mathrm{m/s}$, range 1.1–1.4 m/s). Three-dimensional marker trajectories were captured using a Vicon system (100 Hz, VICON, Oxford Metrics, Oxford, UK) and force data were measured using two AMTI force platforms (1,500 Hz, Advanced Mechanical Technology, Inc., Watertown, MA). A Plug-in-Gait marker set[26] containing lower limb and trunk was used, including a three-marker cluster on both upper and lower legs and one additional medial marker on both knees as well as ankles during the static trials. Thus, a total of 40 markers were included.

Musculoskeletal Modelling and Simulation

The Gait2392 musculoskeletal model installed with OpenSim[27] was used and consisted of 12 segments, 19 degrees of freedom, and 92 musculotendon actuators. Muscle driven simulations and consequent analyses were generated in OpenSim 3.1.[27] The model was first scaled based on the marker locations of a static pose. The scaled model was then used for an inverse kinematics procedure based on measured 3D marker trajectories to determine the kinematics of the movement.[28] Kinematics were low-pass filtered at 6 Hz. Subsequently, a residual reduction algorithm (RRA) was applied, which minimizes the dynamic inconsistency between ground reaction forces and whole body kinematics introduced by errors in modelling and marker kinematics.[29] Two force plates were used to measure force data, therefore, since RRA is only applicable if ground reaction forces under both feet are available, the gait cycle was restricted from left toe off (approx. 12%) until right heel strike (100%).

Using the kinematic solution generated by RRA as a reference, the hip adduction, rotation and flexion angles as well as the pelvis obliquity angle were perturbed by $\pm 5^\circ$ (Fig. 1), both individually and in combination (resulting in 405 simulations, 81 simulations for each subject). Specific joint angle combinations were defined based on gait patterns described in the literature (Table 1).[1,2,4,14] To ensure that for all simulations the ground reaction force location under the foot was identical to the reference simulation, the point of force application was defined in the local reference frame of the foot. As a result, the ground reaction force followed the foot position with varying kinematics.

Changing hip kinematics and adapting the ground reaction forces introduced dynamic inconsistencies between kinematics and external forces. To compensate for these inconsistencies, ideal torque actuators (residual actuators) were applied for each of the six degrees of freedom between pelvis and ground. The torque applied by these actuators represents compensatory behavior that might be obtained by altering trunk kinematics. We, therefore, made two assumptions: (1) ground reaction force direction and magnitude do not change with changing hip kinematics but the point of force application moves with the foot and (2) trunk kinematics is adapted to obtain a dynamically feasible gait pattern, which can be modeled by applying external torques on the pelvis.

Then, for all perturbed simulations, the static optimization procedure as provided in OpenSim[30] was used to calculate muscle forces at each time instant of the gait cycle while minimizing the instantaneous total squared muscle activation. Finally, for all simulations, hip contact forces were calculated using the JointReaction analysis in OpenSim.[31]

Data Analysis

Resultant hip contact forces were expressed in body weight (BW). Furthermore, orientation angles of the hip contact force in the frontal (Ax), transversal (Ay), and sagittal (Az) planes were calculated (Fig. 2).

The resultant hip contact forces showed two peaks, the first peak in early stance, around 17% of the gait cycle and the second peak in late stance, around 50% (Fig. 3). At both instances of the peak contact force, the difference in absolute contact force, orientations, and external joint moments with the reference simulation was calculated. This way the ground reaction force magnitude is equal for each simulation. Therefore, changes in hip contact force result from changes in hip kinematics and kinetics.

To isolate the effect of changing specific kinematic parameters on hip contact forces, the 405 simulations were grouped depending on the perturbed kinematics, resulting in four groups. Each group consists of perturbations in which individual degrees of freedom were perturbed as well as perturbations in which multiple degrees of freedom were simultaneously perturbed and these perturbations are, therefore, present in multiple groups. The first group contained simulations in which the adduction angle was perturbed. The second group contained perturbations of the rotation angle, the third group focused on perturbations of the flexion angle, and the fourth group on perturbations of the pelvis obliquity angle. This resulted in 270 simulations for each of the four groups. Finally, the complete set of 405 simulations was also analyzed.

For each of the grouped simulations, linear and multiple linear regression was used to determine the correlation between the difference in absolute hip flexion, adduction and rotation moments, and the difference in absolute hip contact force at the time of the first and second peak. Coefficients of determination (R^2) as well as regression equations were

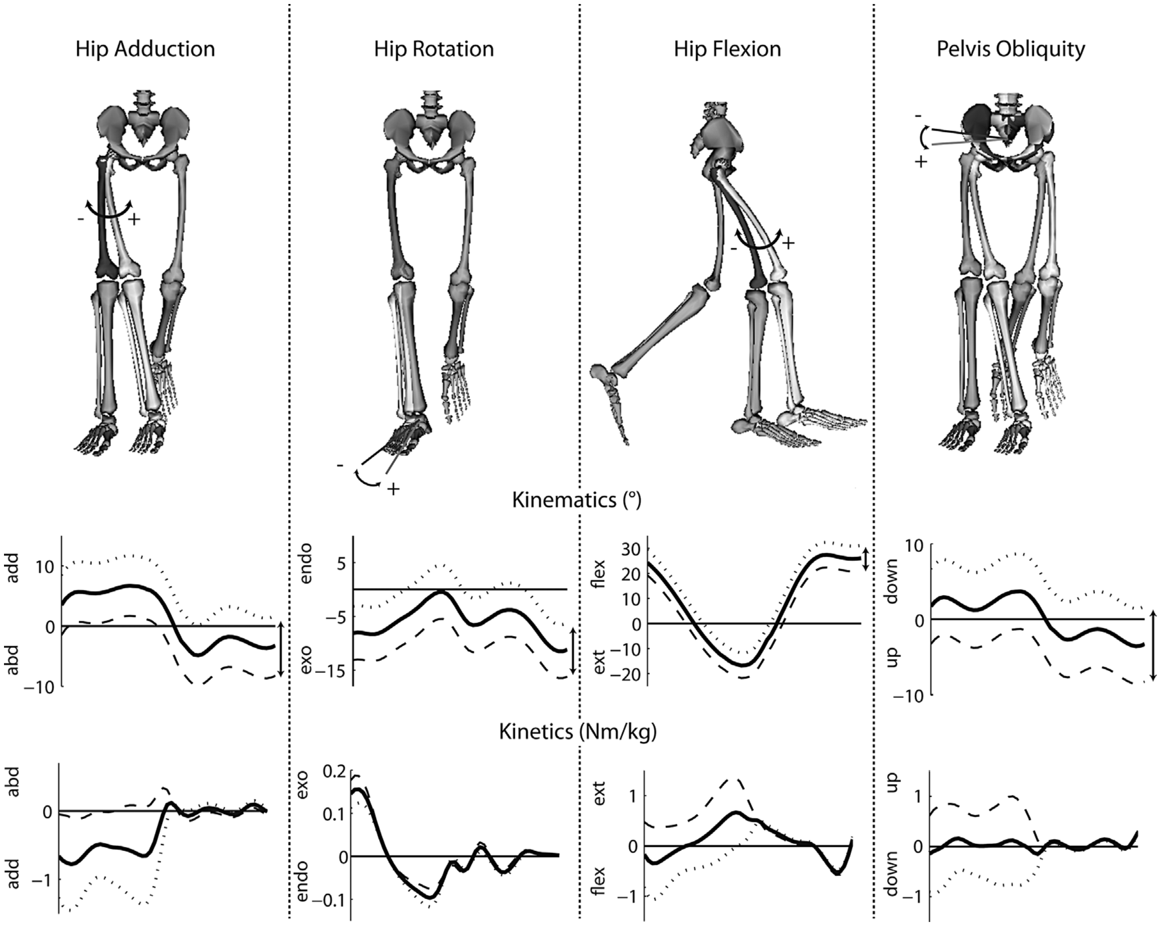

Figure 1. The different perturbations by ±5° in hip and pelvis kinematics. Specifically, the hip adduction, rotation, and flexion angles as well as pelvis obliquity were perturbed. The kinematic graphs show the average nominal joint angles (solid line) and the perturbations by +5° (dotted line) and −5° (dashed line). The kinetic graphs show the average nominal joint moments (solid line) and the moments when perturbing the kinematics by +5° (dotted line) and −5° (dashed line).

calculated. R^2 in the range 0.5–0.7 were considered to be low, in the range 0.7–0.9 moderate, and above 0.9 high.[32] The results were significant when $p < 0.05$. R^2 values were calculated for the complete set of 405 simulations as well as for the four different subgroups. R^2 between the orientation angles and hip moments were calculated only for the complete set of simulations.

Table 1. Gait Patterns Described in Hip Pathology Patients and the Perturbations That Were Done to Simulate These Gait Patterns

	Gait Pattern	Implementation
First peak	Decreased hip adduction	adduction −5°
	Decreased pelvis obliquity	obliquity −5°
	and increased exorotation	rotation −5°
	Trendelenburg	adduction +5°
		obliquity −5°
Second peak	Decreased hip extension	extension −5°
	Decreased pelvis obliquity	adduction +5°
	with increased hip	rotation −5°
	exorotation and adduction	obliquity −5°
	Trendelenburg	adduction +5°
		obliquity −5°

RESULTS

Effect on Peak Hip Contact Force

The first (Fig. 4) and second peak (Fig. 5) in hip contact force were similarly influenced by most kinematic perturbations. Changes in frontal plane hip and pelvis kinematics had the largest effect on the magnitude of hip contact forces. Decreased hip adduction and ipsilateral pelvic drop (decreased pelvic obliquity) decreased hip contact forces, whereas increased hip adduction and increased pelvic obliquity had an opposite effect. These changes in hip contact forces could be related to the concomitant changes in hip adduction moment. Results for the orientation of the hip contact force vector are reported in supplementary material A.

The effect of hip flexion on hip contact forces was smaller. Both increased and decreased hip flexion increased the first peak hip contact force (Fig. 4). At the second peak, more hip extension led to increased contact forces, while a reduced hip extension led to decreased contact forces (Fig. 5).

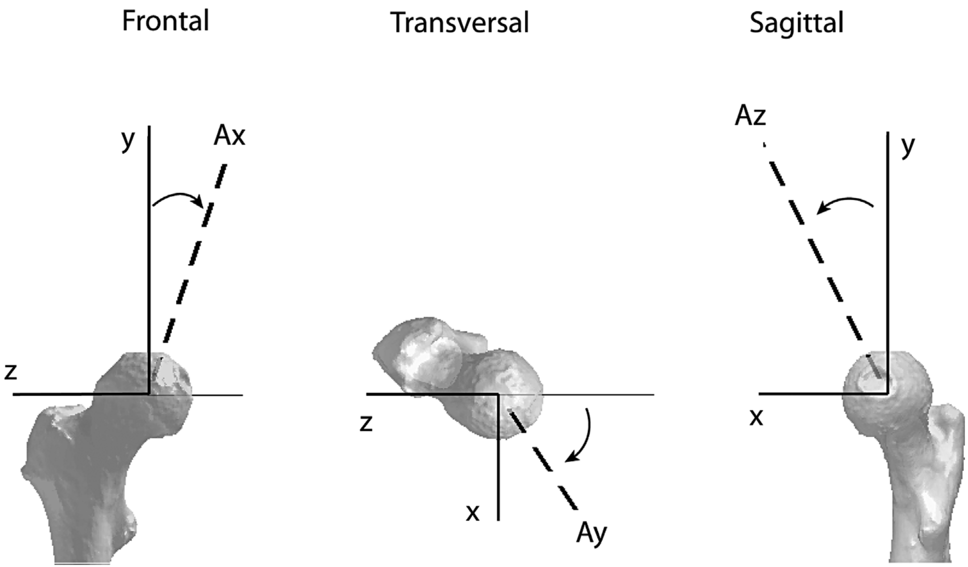

Figure 2. The calculated orientation angles in the frontal (Ax), transversal (Ay), and sagittal (Az) planes.

Changes of the transversal plane kinematics had a small effect on hip contact forces. Decreased hip rotation resulted in increased contact forces, while increased rotation resulted in decreased contact forces at the first peak (Fig. 4). At the second peak, results were opposite and smaller (Fig.5).

All clinical gait patterns (Table 1) showed that hip contact forces and moments were decreased both at the first and second peak (Figs. 4 and 5), although for the Trendelenbug gait (adduction $+5^{\circ}$ with obliquity -5°) differences were small.

Correlation Between Hip Moment and Hip Contact Force

At the first peak, good correlations of the hip contact forces with the hip adduction and rotation moments separately were found. Coefficients of determination (R^2) were similar for the separate subgroups (Table 2) as for the complete set of simulations (Table 3). When multiple hip moments in different planes were taken into account, there was only a small increase in correlation with the hip contact force (Table 3).

At the second peak, coefficients of determination with any of the separate hip moments were low. Hip

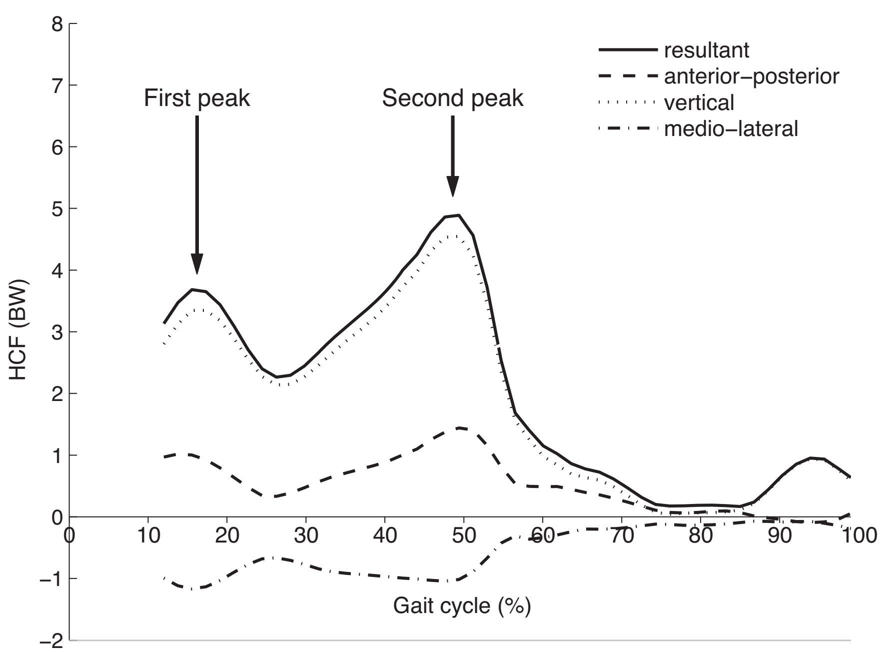

Figure 3. Average nominal hip contact force (normalized to body weight [BW]) shown from toe off of the left leg until heel strike of the right leg. The resultant contact force as well as the different force components are shown. Two peaks were defined, i.e., the first (at 15–20% of the gait cycle) and second peak (at 45–55% of the gait cycle) in hip contact force.

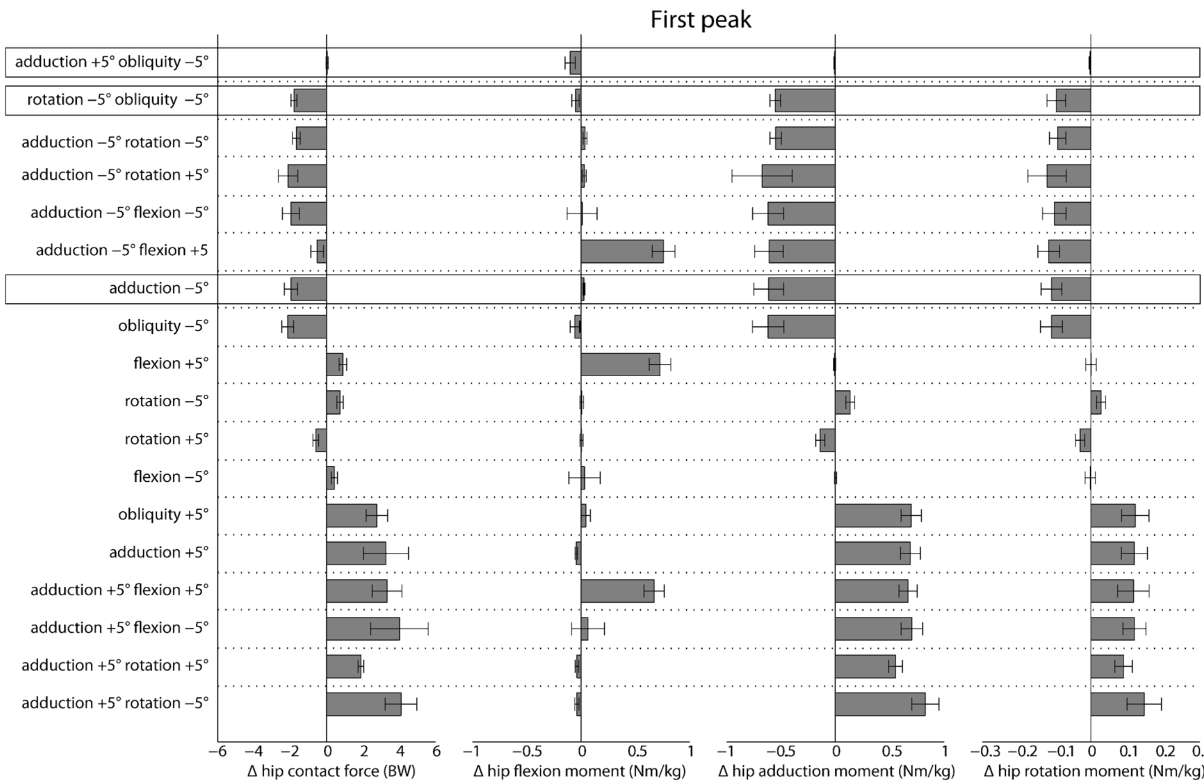

Figure 4. Difference with the nominal simulation in hip contact forces and hip flexion, adduction, and rotation moments at the first peak for different kinematic perturbations. The difference in absolute contact force and moments with the nominal simulations is shown. The squares indicate the gait patterns described in hip pathology patients.

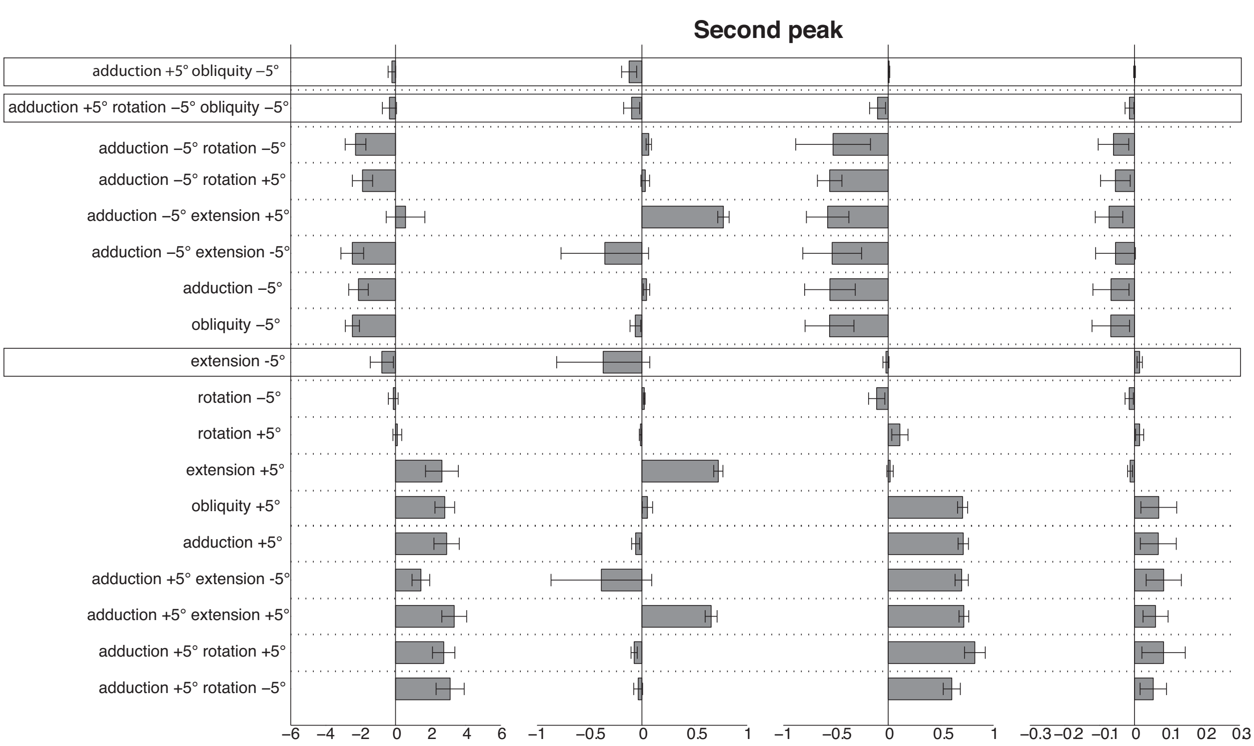

Figure 5. Difference with the nominal simulation in hip contact forces and hip flexion, adduction, and rotation moments at the second peak for different kinematic perturbations. The difference in absolute contact force and moments with the nominal simulations is shown. The squares indicate the gait patterns described in hip pathology patients.

Table 2. Coefficients of Determination (R^2), Including p Values, of the Hip Contact Forces With Hip Flexion, Adduction, and Rotation Moments, Respectively, at the Occurrence of the Peaks in Hip Contact Force. Four Different Subgroups are Defined. Each Group Consists of Perturbations in Which Individual Degrees of Freedom (DOF) Were Perturbed as Well as Perturbations in Which Multiple DOF Were Simultaneously Perturbed and These Perturbations are, Therefore, Present in Multiple Groups. This Resulted in 270 Simulations for Each of the Four Groups. Moderate and High Correlations are Printed in Bold

	Perturbed DOF	Flexion Moment	Adduction Moment	Rotation Moment
First peak	Adduction	0.01 $p = 0.093$	**0.87 $p < 0.001$**	**0.84 $p < 0.001$**
	Rotation	0.02 $p = 0.020$	**0.87 $p < 0.001$**	**0.86 $p < 0.001$**
	Flexion	0.01 $p = 0.084$	**0.88 $p < 0.001$**	**0.84 $p < 0.001$**
	Obliquity	0.02 $p = 0.025$	**0.88 $p < 0.001$**	**0.86 $p < 0.001$**
Second peak	Adduction	0.16 $p < 0.000$	0.54 $p < 0.000$	0.32 $p < 0.000$
	Rotation	0.20 $p < 0.001$	0.54 $p < 0.001$	0.31 $p < 0.001$
	Flexion	0.30 $p < 0.001$	0.44 $p < 0.001$	0.23 $p < 0.001$
	Obliquity	0.20 $p < 0.001$	0.55 $p < 0.001$	0.33 $p < 0.001$

flexion and adduction moment combined showed a moderate R^2, whereas adding the rotation moment did not further increase R^2 (Table 3). Results of the orientation are again reported in supplementary material A.

DISCUSSION

The goal of the study was (1) to investigate to what extent hip and pelvis kinematics can affect hip contact forces and (2) to define gait patterns that can be used to influence the hip contact forces. The results showed that mainly a change in frontal plane hip and pelvis kinematics had an effect on the magnitude of hip contact forces, where the effect of changes in ipsilateral pelvic drop (pelvic obliquity) was similar to the effect of changes in hip adduction. To decrease hip contact forces, the hip adduction angle should be decreased, which may result in a more wide based gait pattern. On the other hand, a narrower base of support due to an increase in hip adduction will effectively increase the hip contact forces.

Results showed that the changes in hip contact forces were related to changes in hip moments. More specifically, high coefficients of determination for the adduction moment were found at the first peak contact forces (Table 3), both for the complete set of simulations as well as for the separate kinematic subgroups (Table 2). This indicates that the hip adduction moment can predict the contact forces well, independent of the specific kinematic strategy. Different combinations of kinematic perturbations can result in a similar change in hip moments and therefore, a similar change in hip contact forces. The relation between hip adduction angle and hip contact force is additionally investigated, which showed that the hip adduction angle in isolation cannot be used to predict the change in hip contact force (supplementary material B). Therefore, the change in hip adduction moment,

Table 3. Regression Equations and Coefficients of Determination (R^2), Including p Values, of the Hip Contact Forces with Hip Flexion, Adduction, and Rotation Moments Both Individually as Well as for Combined Hip Moments at the Occurrence of the Peaks in Hip Contact Force. The Complete Set of 405 Simulations is Taken Into Account. Moderate and High Correlations are Printed in Bold

	Hip Moment	Regression Equation	R^2
First peak	Flexion	$\Delta HCF = 1.02{*}\Delta M_{flex} + 0.80$	0.02 $p = 0.007$
	Adduction	$\Delta HCF = 3.85{*}\Delta M_{add} + 0.47$	**0.87 $p < 0.001$**
	Rotation	$\Delta HCF = 20.31{*}\Delta M_{rot} + 0.60$	**0.85 $p < 0.001$**
	Flexion adduction	$\Delta HCF = 1.00{*}\Delta M_{flex} + 3.85{*}\Delta M_{add} + 0.23$	**0.89 $p < 0.001$**
	Adduction rotation	$\Delta HCF = 2.68{*}\Delta M_{add} + 6.50{*}\Delta M_{rot} + 0.51$	**0.88 $p < 0.001$**
	Rotation flexion	$\Delta HCF = 20.27{*}\Delta M_{rot} + 0.93{*}\Delta M_{flex} + 0.37$	**0.87 $p < 0.001$**
	Rotation flexion adduction	$\Delta HCF = 6.10{*}\Delta M_{rot} + 2.75{*}\Delta M_{add} + 0.98{*}\Delta M_{flex} + 0.26$	**0.90 $p < 0.001$**
Second peak	Flexion	$\Delta HCF = 2.04{*}\Delta M_{flex} + 0.46$	0.20 $p < 0.001$
	Adduction	$\Delta HCF = 2.75{*}\Delta M_{add} + 0.15$	0.54 $p < 0.001$
	Rotation	$\Delta HCF = 17.70{*}\Delta M_{rot} + 0.37$	0.32 $p < 0.001$
	Flexion adduction	$\Delta HCF = 2.12{*}\Delta M_{flex} + 2.79{*}\Delta M_{add} - 0.10$	**0.76 $p < 0.001$**
	Adduction rotation	$\Delta HCF = 2.95{*}\Delta M_{add} - 2.06{*}\Delta M_{rot} + 0.15$	0.55 $p < 0.001$
	Rotation flexion	$\Delta HCF = 19.06{*}\Delta M_{rot} + 2.29{*}\Delta M_{flex} + 0.08$	0.57 $p < 0.001$
	Rotation flexion adduction	$\Delta HCF = 0.79{*}\Delta M_{rot} + 2.71{*}\Delta M_{add} + 2.13{*}\Delta M_{flex} - 0.10$	**0.76 $p < 0.001$**

rather than kinematics should be used as feedback to evaluate the change in hip contact force. Foucher et al.[16] found only moderate correlations between the hip moments and contact forces. In contrast to the present study, they related peak joint moments to the first and second peak contact forces.

At the second peak, lower R^2 values were found (Tables 2 and 3). However, combining the hip adduction and flexion moments increased R^2 at this time instant. This indicates that both the hip adduction and flexion moments are required to accurately predict hip contact forces at the second peak. A decreased coefficient of determination in late stance has also been reported previously by Kutzner et al. for the knee during late stance.[8]

The effect of gait patterns described in hip pathology patients on joint contact forces has also been investigated. A decreased hip adduction angle and moment have been reported previously in patients after THA.[1] At the first peak in contact force, this gait pattern resulted in decreased contact forces (Fig. 4) and a more vertical loading (supplementary material A). At the first peak, decreased pelvis obliquity in combination with decreased rotation has also been reported and linked to decreased hip contact forces in patients after THA.[2] This combination of parameter perturbations also resulted in decreased contact forces in the current study (Fig. 4) and a more medial and vertical loading (supplementary material A).

In contrast, a Trendelenburg gait, i.e., increased hip adduction and decreased pelvis obliquity, did not importantly affect the contact forces compared to the reference simulation, indicating that this gait pattern does not result in modified loading of the hip joint. At the second peak, a small decrease in contact force was found (Fig. 5). This suggests that a Trendelenburg gait does not induce excessive hip contact forces at either the first or second peak. However, the orientation of the hip contact force was more anterior and vertical at the first peak, while at the second peak a more medial loading was found (supplementary material A).

Another gait adaptation that has been reported is a decrease in hip extension at terminal stance.[1,4] This resulted in decreased contact forces, although to a lesser extent than for hip adduction angle alterations (Fig. 5), and a more medial and vertical loading (supplementary material A). Lenaerts et al.[2] reported a significant decrease in downward pelvis obliquity and endorotation together with increased hip adduction, which resulted in a not significant increase in hip contact force. However, in the present study this combination of kinematic perturbations resulted in slightly decreased contact forces and more medial and anterior loading of the hip. The decreased downward obliquity reported by Lenaerts et al. was smaller than the increased hip adduction, which can explain the slight increase in hip contact forces. Also, significant changes in pelvis rotation and knee flexion were found, but these angles were not accounted for in the present study.

The decreased loading that is associated with the gait patterns observed in hip pathology patients has several potential benefits. Decreased loading enhances bony ingrowth of the prosthesis in the first weeks after surgery.[33] Furthermore, increased joint loading has been defined as a risk factor for developing OA.[34] Most often, patients are trained to return to a normal gait pattern while the effect on hip contact forces is not considered. Gait retraining that focuses on decreasing hip adduction (i.e., wide base of support) can be considered to induce a decreased joint loading. The effect of specific alterations in the orientation of the hip contact force is, however, not yet defined.

Apart from the decreased hip contact forces, the effect on the other joints should be considered as well: At the lumbar joint, changes in contact force are low (supplementary material C). At the knee joint the effect is larger: With increasing hip adduction contact forces at the knee are increased, decreasing hip adduction resulted in decreased knee contact forces, although less pronounced. This indicates that the kinematic changes that decrease hip contact forces do not result in overloading in the knee joint.

A limitation of the present study is that the dynamic inconsistencies between the kinematics and external forces are compensated by the use of residual actuators. It is assumed that trunk kinematics are adapted to obtain a dynamically feasible gait pattern. However, other compensation strategies can be adopted, which can affect the direction and magnitude of the ground reaction force. This will have an effect on the hip moments and therefore contact forces, and should be further investigated in future research. Further, a generic musculoskeletal model has been used. However, the current modelling pipeline did not include subject-specific bone and muscle geometry, as no medical imaging data was available. The use of subject-specific models can affect calculated contact forces.[35,36] Also, a static optimization was used to calculate muscle forces. This method simplifies the muscle physiology and does not account for muscle dynamics. Also, all subjects were lean, therefore, the effect on more obese subjects should be further investigated. Besides that, five subjects were used to define the average baseline data. However, a larger number of subjects will result in a dataset that is more representative for a larger population. Also, the joint angles were perturbed over $\pm 5°$, while larger differences with control subjects have been found. Therefore, the reported results are only applicable for small changes in joint angles ($\pm 5°$). The relation for larger perturbations in the population of hip pathology patients should be further investigated.

Average nominal first and second peak hip contact forces (3.68 BW and 4.89 BW, respectively) were higher than measured using instrumented prostheses[37] (2.33 BW and 2.05 BW). This overestimation has been described before,[24,38] although contact forces closer to measured forces have also been reported.[39,40]

This might be attributed to different modelling choices (e.g., the use of a generic musculoskeletal model) or differences in subject and gait characteristics.[38] However, the modeling choices do not affect the comparisons presented in this paper, since all simulations were based on the same model.

In conclusion, changes in hip and pelvis kinematics resulted in changes in hip kinetics which caused a change in hip contact forces. Specifically, a decreased hip adduction, i.e., a wide base gait pattern, decreased contact forces. The clinical gait patterns resulted in decreased hip contact forces, indicating that hip patients adopt their gait pattern to decrease the loading on the hip. However, a Trendelenburg gait did not increase the joint contact forces. Furthermore, contact forces correlated well with specifically the hip adduction and rotation moments in early stance. In late stance, the combined hip adduction and flexion moments were needed to predict the contact forces. These results suggest that gait training that focuses on decreasing hip adduction angles and resulting moments may reduce hip contact forces.

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