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Using musculoskeletal modelling to estimate knee joint loading pre and post high tibial osteotomy

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ABSTRACT

Background: Both medial knee osteoarthritis and associated varus alignment have been proposed to alter knee joint loading and consequently overloading the medial compartment. Individuals with knee osteoarthritis and varus deformity are candidates for coronal plane corrective surgery, high tibial osteotomy. This study evaluated knee loading and contact location for a control group, a pre-surgery cohort and the same cohort 12 months post-surgery using a musculoskeletal modelling approach.

Methods: Joint kinematics during gait were measured in 30 knee osteoarthritis patients, before and after high tibial osteotomy, and 28 healthy adults. Using a musculoskeletal model that incorporated patient-specific mechanical tibial femoral angle, the resulting muscle, ligament, and contact forces were calculated and the medial - lateral condyle load distribution was analysed.

Findings: Surgery changed medial compartment contact force throughout stance relative to pre-surgery. This reduction in medial compartment contact force pre- vs post-HTO is observed despite a significant increase in post-surgery walking speed compared to pre-HTO, where increased speed is typically associated with increased joint loading.

Interpretation: This study has estimated the effects of high tibial osteotomy on knee loading using a generic model that incorporates a detailed knee model to better understand tibiofemoral contact loading. The findings support the aim of surgery to unload the medial knee compartment and lateralise joint contact forces.

1. Introduction

The magnitude and location of joint contact forces are important to consider when assessing the causes and treatment of knee pathologies (Andriacchi et al., 2004). Varus alignment has been proposed to alter knee joint loading and is associated with medial knee osteoarthritis (mKOA) (Whatling et al., 2019), overloading the medial compartment. Furthermore, overloading of the medial knee compartment has been strongly associated with KOA progression (Miyazaki et al., 2002) and radiographic disease severity (Sharma et al., 1998).

Individuals with KOA and varus deformity are candidates for coronal plane corrective surgery, high tibial osteotomy (HTO). HTO aims to unload the medial compartment by lateralising the weight bearing line (Jackson et al., 1969) and thus shifting joint contact forces with the goal to reduce KOA progression.

Our research group has shown that HTO surgery restores frontal plane loading as reflected in external joint moments. First peak external knee adduction moment (EKAM) group mean value changed from 3.02 %BW pre-HTO to 2.06 %BW.h at 12 months post-surgery (Whatling et al., 2019). However, whether an increased EKAM adequately represents increased medial compartment tibiofemoral loading remains a subject for debate. Internal joint loads cannot normally be measured in-vivo, however computational models can be employed to estimate the joint contact forces arising during functional tasks, such as walking, to better understand the effect of surgical interventions (Lenhart et al., 2015). The study by Bhatnagar and Jenkyn (2010) used a computational knee model to estimate internal kinetics during walking gait pre- and post-

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HTO. First peak of an EKAM during stance phase was reduced significantly by surgery. The medial compartment load and medial-to-lateral compartment loads were also significantly reduced. The authors concluded that the external measure EKAM-impulse is a good proxy of the internal kinetic measure of a medial-to-lateral compartment loads ratio-impulse.

A model-based study by two authors of this paper (van Rossom et al., 2019) simulated altered tibiofemoral alignment and found that coronal plane knee malalignment affects knee loading; with increased varus alignment resulting in increased medial loading. This confirms a causal relation between coronal malalignment and increased medial compartment loading and suggests a role of aberrant coronal plane alignment on KOA initiation (van Rossom et al., 2019). However, van Rossom et al. (2019) studied the alterations in a healthy cohort and so neglected KOA-specific movement patterns. It is of clinical importance to determine whether normal medial compartment knee contact loading is restored following HTO. This can provide valuable evidence for surgical efficacy in terms of lateralising knee loading with the potential to delay disease progression and thereby prolonging the need for further surgical interventions such as a total knee replacement. More recently, De Pieri et al. (2022) published work on how HTO effectively redistributes compressive knee loads during gait. Musculoskeletal modelling was used to evaluate lower limb joint moments and knee contact forces during gait. De Pieri et al. (2022) concluded that HTO effectively unloaded the medial compartment by redistributing part of the overall compressive force to the lateral compartment during gait with limited influence on gait function.

The purpose of this study is to use musculoskeletal modelling to estimate medial compartmental knee loading and contact location for a control group, a pre-HTO surgery KOA patient cohort and the same HTO cohort 12 months post-surgery. It is hypothesised that higher medial condyle tibiofemoral joint contact forces would be observed in the KOA cohort pre-HTO, whereas the magnitude and position of the knee forces would become similar to the control group at 12 months post-surgery.

2. Methods

2.1. Participants

29 participants (30 knees) with KOA and varus alignment were recruited from the out-patient clinic of the local senior surgeon (CW). Approval for this work was granted by the Wales Research Ethics Committee 3 (10/MRE09/28) and Cardiff and Vale University Health Board. Written informed consent was obtained from each participant prior to data collection.

Patients were included if they were between the ages of 18 and 80 and listed for medial opening wedge HTO. Patients did not pass initial screening if they were unable to provide informed consent, had neurological or visual conditions affecting movement or a previous injury to the joint under investigation that the treating clinician deemed unsuitable. The extent of KOA was determined using the Kellgren–Lawrence (KL) (Kellgren and Lawrence, 1957) radiographic score and varus alignment calculated as the mechanical tibiofemoral angle (mTFA) from long leg weight bearing radiographs.

Medial opening wedge HTO surgery was used to correct varus deformity using standard surgical approaches and planning (Brinkman et al., 2008). The osteotomies were fixed with either Tomofix ($n = 27$) or Puddu ($n = 1$) plates, shown to have similar biomechanical properties (Golovakha et al., 2014), or an iBalance device (n = 2).

28 subjects with no lower limb pathology were recruited from university staff, students and community using advertisements, forming a non-pathological control cohort (controls). Microsoft Excel was used to randomly assign control participant to either left or right to randomly distribute limb allocation.

2.2. Motion analysis

Three-dimensional gait analysis was performed on patients before (average 1.4 ± 1.4 months) and after HTO surgery (average 13 ± 3.4 months) and at one time point for the control group. Gait analysis was performed using an 8 or 12 Oqus camera system (Qualisys, Sweden) capturing at 120 Hz, synchronised with either two, four or six (due to laboratory upgrades) force platforms (Bertec Corp., USA) capturing at either 1080 Hz or 2000 Hz. Markers were placed following a modified Cleveland markerset, as implemented in previous publications (Whelton et al., 2017; Whatling et al., 2019). Subjects walked at their self-selected speed for a minimum of 3 successful trials.

2.3. Simulation framework

Data was processed using a previously validated musculoskeletal modelling workflow (Lenhart et al., 2015) to obtain the tibiofemoral contact forces and pressures (Lenhart et al., 2015). It integrates an extended knee model, that allows 6 degrees of freedom (DoF) patellofemoral and tibiofemoral movement, in a generic full-body model (Lenhart et al., 2015). Each leg included 44 musculotendon actuators spanning the hip, knee, and ankle and 14 bundles of non-linear springs that represent the major knee ligaments and posterior capsule. A non-linear elastic foundation formulation was used to calculate the cartilage contact pressures, based on the penetration depth of the overlapping surface meshes of the contact model (Smith et al., 2018). The cartilage was modelled with a uniformly distributed thickness of 4 mm tibiofemoral and 7 mm patellofemoral thickness (Eckstein et al., 2001; Hudelmaier et al., 2003). The elastic modulus and Poisson's ratio were assumed as 10 MPa and 0.45, respectively (Adouni and Shirazi-Adl, 2014; Li et al., 2001). This model was implemented in SIMM with the Dynamics Pipeline (Musculographics Inc., Santa Rosa, CA) and SD/Fast (Parametric Technology Corp., Needham, MA) to generate the multibody equations of motion.

At first, the generic model was scaled to the subjects' anthropometry, where the patient-specific mechanical tibiofemoral angle (mTFA) was implemented in the model, comparable to van Rossom et al. (2019). To this end, the tibia geometry was rotated to simulate the malalignment of the affected lower limb. When simulating with the malalignment, the location of the foot with respect to the measured GRF application point was medialised. To ensure that for these simulations the application of the GRF to the foot was identical to the reference simulation, the COP (of the GRF) was expressed in the local reference frame of the foot. Varus alignment was accounted for by using the mTFA angle and implemented within the original COMAK model. When doing this the configuration of the bone file, joint file, contact file, muscle file, and muscle-tendon parameters were accounted for.

The model was altered so that, in a static pose, the hip-knee-ankle angle (frontal plane) better matched the patients lower limb alignment. In simpler models, researchers have implemented a knee adduction angle to OpenSim models for scaling, then locked this DOF for dynamic activities. The reasoning behind this is that bone pin data (Benoit et al., 2006) shows that the knee moves very little in frontal plane during gait.

For the Lenhart model, the tibiofemoral adduction is defined by the geometry of the contact surfaces and so it was not feasible to manually alter this. Therefore, it was decided to adjust the tibial and foot segments, i.e., ankle joint. The ankle is translated to a new location based on the mTFA angle to effectively introduce varus/valgus and corrected for foot alignment, so it lands flat on the floor.

Next, joint angles (pelvic translations and rotations, hip flexion, hip adduction, hip rotation, knee flexion and ankle flexion) were calculated using inverse kinematics. Subsequently, the muscle forces and secondary knee kinematics (11 DoF, i.e., all except knee flexion) required to generate the measured primary hip, knee and ankle accelerations were estimated using the concurrent optimisation of muscle activations and kinematics algorithm. In the optimisation the weighted sum of squared muscle activations and contact energy were minimised (Smith et al., 2018). As only the knee flexion angle was used in the optimisation, joint kinematics in the secondary knee DoF evolved as a function of muscle, ligament, and contact forces (Lenhart et al., 2015; Smith et al., 2018).

2.4. Study statistics

For each trial, the stance phase was identified as the period in which the ground reaction force (GRF) exceeded 20 N. This study averaged 3 trials for each participant.

The magnitude and timing of the first and second peak (FP and SP) of the resultant tibiofemoral contact force was determined during the first and second half of the stance phase, respectively, as well as the minimum force during single leg support (MS). Each variable was determined for the FP, MS, and SP for total knee and were averaged over three trials. The concomitant average and maximum pressure over the contact surface was analysed. Furthermore, the point of application of the total knee compartment, medial knee compartment and lateral knee compartment contact force expressed in the local reference frame of the tibia at FP, SP and MS were analysed.

Paired samples *t*-test was performed in MATLAB (MathWorks, USA) to identify significant differences associated with HTO surgery. Where parametric assumptions were not met, a Wilcoxon signed-rank test was used. Independent *t*-tests were used to determine significant differences in the pre- and post-HTO measurements compared to the control group. Where parametric assumptions were not met, a Mann–Whitney *U* test was performed. Significance was determined when $p < 0.05$ for all statistical tests.

3. Results

Table 1 shows participant demographics and clinical measures. Post-surgery, patients walked with a faster gait speed compared to pre-surgery ($P = 0.020$) but remained slower than the control group. Patients spent longer in the stance phase compared to the controls both pre- and post-HTO ($P < 0.001$).

Table 1
Group characteristics.

Demographics	Controls Mean (std)	Pre-HTO Mean (std)	Post-HTO Mean (std)	Controls vs pre-HTO *P* value	Controls vs post-HTO *P* value	Pre vs Post HTO *P* value
Number of knees	28	30	30			
Gender (M/F)	13/15	25/5	25/5			
Mean age, years (SD)	38.14 (11.09)	50.70 (8.71)	51.83 (8.79)	**<0.001**††	**<0.001**††	
Height (m)	1.70 (0.08)	1.75 (0.11)	1.75 (0.11)	**0.018***	**0.017**	
Mass (kg)	71.84 (15.74)	90.57 (20.17)	89.92 (19.98)	**<0.001**††	**<0.001**††	0.242
BMI (kg/m^2)	24.96 (4.36)	29.27 (5.04)	29.08 (4.93)	**0.001**††	**0.001**††	0.302
KL Grade	n/a	6 KL2; 20 KL3; 4 KL4				
mTFA (°)	n/a	7.75 (3.72) varus	0.92 (2.82) varus (n = 27)			**<0.001****
Stance time (s)	0.69 (0.05)	0.78 (0.12)	0.76 (0.08)	**<0.001**††	**<0.001**††	0.336
Gait speed (m/s)	1.26 (0.17)	1.1 (0.24)	1.15 (0.17)	**0.006****	**0.020***	**0.020***

Signifcant difference (p < 0.01) indicated by ** where parametric or †† where non-parametric tests used.

3.1. Knee loading

3.1.1. Contact forces

Pre-surgery, patients had lower total knee contact forces compared to control subjects throughout the gait cycle (Table 2, Fig. 1); possibly related to the significant differences found in gait speed between the KOA and control cohorts. Notwithstanding the differences in contact forces, patients' pre-surgery had significantly higher average and maximum articular cartilage pressures in both the total and medial compartment of the tibiofemoral joint during midstance compared to the control cohort (Fig. 2). In the latter part of stance, patients' pre-surgery had significantly lower lateral knee contact force, but significantly higher lateral knee mean pressure and maximum pressure values compared to the control cohort.

Surgery resulted in significant reductions in medial compartment contact forces and mean pressure throughout the first half of stance (Tables 2 and 3; Figs. 1 and 2). Additionally, reductions were observed in total knee maximum pressure during midstance (9.97 MPa (2.03) vs 8.93 MPa (1.59), $P < 0.001$) due to the pressure reduction in the medial compartment (9.73 MPa (2.07) vs 8.59 MPa (1.51), $P = 0.001$).

At 12 months post-HTO, there was a significant decrease in medial compartment contact force in the KOA group compared to the controls at both the first half of stance (1.69 BW (0.31) vs 1.46 BW (0.36), $P = 0.011$) and second half of stance (1.83 BW (0.35) vs 1.51 BW (0.37), $P = 0.001$). Surgery resulted in no significant differences between the control cohort and post-HTO in total knee contact force during midstance (1.25 BW (0.24) vs 1.23 BW (0.18), $P = 0.86$). Additionally, when comparing the control cohort to 12-months post-HTO, there were no significant differences in lateral compartment knee contact force (1.2 BW (0.44) vs 1.01 (0.3), $P = 0.071$) in the second half of stance compared to there being significant differences between the control cohort and pre-HTO.

3.1.2. Tibiofemoral loading location

Pre-surgery, KOA patients presented with a more medialised location of the point of application within the medial knee compartment compared to control subjects throughout the whole of stance (Table 3), and a more posterior contact force location within the lateral knee compartment compared to the control cohort from midstance to toe-off.

HTO surgery resulted in significant move from a more medial to lateral position within knee compartments throughout stance (Table 3; Fig. 2).

Point of force application metrics were not significantly different when comparing 12 months post-HTO to the control cohort, except for those representing the second peak in the medial and lateral compartments which were more lateral (Table 3).

3.1.3. Contact area

Pre-surgery, patients presented with a larger medial compartment contact area compared to the control cohort at both the first peak (206.8 mm^2 (22.63) vs 223.16 mm^2 (29.08), $P = 0.012$) and at midstance (167.14 mm^2 (24.46) vs 223.16 mm^2 (29.08), $P = 0.012$) (Table 4; Fig. 3).

Surgery resulted in a significant decrease in medial compartment contact area (223.16 mm^2 (29.08) vs 213.55 mm^2 (32.49), $P = 0.030$) and a significant increase in lateral compartment contact area (139.1 mm^2 (29.50) vs 149.43 mm^2 (21.45), $P = 0.030$) (Table 4).

12 months post-HTO resulted in normalisation of all contact area metrics between the KOA group compared to the controls (Table 4).

Table 2
Tibiofemoral loading variables.

	Controls	Pre-HTO	Post-HTO	Controls vs pre-HTO	Controls vs post-HTO	Pre vs Post HTO
	Mean (std)	Mean (std)	Mean (std)	*P* value	*P* value	*P* value
First peak						
Total knee						
Contact force [BW]	2.74 (0.57)	2.41 (0.57)	2.33 (0.52)	**0.033***	**0.006****	0.418
Mean pressure [MPa]	5.63 (1.25)	6.11 (1.63)	5.77 (1.13)	0.349	0.671	0.144
Max pressure	12.92 (3.32)	14.19 (4.13)	13.22 (2.46)	0.248	0.501	0.094
Medial knee						
Contact force [BW]	1.69 (0.31)	1.61 (0.36)	1.46 (0.36)	0.362	**0.011***	**0.017***
Mean pressure [MPa]	5.76 (1.12)	6.44 (1.67)	5.95 (1.12)	0.073	0.526	**0.046***
Max pressure	12.21 (2.52)	13.36 (3.72)	12.69 (2.45)	0.171	0.469	0.184
Lateral knee						
Contact force [BW]	1.13 (0.37)	0.87 (0.38)	0.94 (0.27)	**0.011***	**0.027***	0.332
Mean pressure [MPa]	5.37 (1.62)	5.38 (2.12)	5.45 (1.43)	0.969	0.883	0.837
Max pressure	11.57 (3.76)	11.48 (4.71)	11.78 (3.12)	0.908	0.671	0.677
Midstance						
Total knee						
Contact force [BW]	1.25 (0.24)	1.36 (0.25)	1.23 (0.18)	**0.037***	0.859	**0.001††**
Mean pressure [MPa]	3.4 (0.36)	4.36 (0.81)	3.94 (0.58)	**<0.001††**	**<0.001****	**0.001****
Max pressure	7.62 (1.12)	9.97 (2.03)	8.93 (1.59)	**<0.001††**	**0.001****	**<0.001****
Medial knee						
Contact force [BW]	0.92 (0.22)	1.03 (0.26)	0.89 (0.2)	0.084	0.644	**0.001****
Mean pressure [MPa]	3.81 (0.58)	4.79 (0.91)	4.31 (0.7)	**<0.001****	**0.004****	**0.001****
Max pressure	7.43 (1.22)	9.73 (2.07)	8.59 (1.51)	**<0.001****	**0.002****	**0.001****
Lateral knee						
Contact force [BW]	0.36 (0.14)	0.36 (0.21)	0.37 (0.15)	0.920	0.660	0.911
Mean pressure [MPa]	2.59 (0.54)	3.03 (1.37)	3.04 (1.06)	0.255	0.070	0.530
Max pressure	5.56 (1.07)	6.29 (2.85)	6.42 (2.17)	0.333	0.065	0.772
Second peak						
Total knee						
Contact force [BW]	2.9 (0.7)	2.48 (0.62)	2.43 (0.54)	**0.018***	**0.006****	0.621
Mean pressure [MPa]	5.32 (0.64)	5.98 (1.08)	5.77 (1.11)	**0.022†**	0.147	0.201
Max pressure	12.7 (1.76)	14.12 (3.05)	13.34 (3.05)	0.126	0.811	0.058
Medial knee						
Contact force [BW]	1.83 (0.35)	1.63 (0.47)	1.51 (0.37)	0.075	**0.001****	0.100
Mean pressure [MPa]	5.91 (0.8)	6.42 (1.36)	6.14 (1.23)	0.084	0.386	0.263
Max pressure	12.61 (1.8)	13.14 (2.83)	12.65 (2.76)	0.393	0.605	0.333
Lateral knee						
Contact force [BW]	1.20 (0.44)	0.94 (0.37)	1.01 (0.3)	**0.021***	0.071	0.148
Mean pressure [MPa]	4.53 (0.73)	5.2 (1.48)	5.23 (1.39)	**0.045†**	**0.014†**	0.889
Max pressure	9.43 (1.48)	11.21 (3.61)	11.2 (3.3)	**0.021†**	**0.003††**	0.991

Significant difference ($p < 0.01$) indicated by ** where parametric or †† where non-parametric tests used.

4. Discussion

The purpose of this study is to use musculoskeletal modelling to estimate medial compartmental knee loading and contact location for a control group, a pre-HTO surgery KOA patient cohort who had varus deformity of the lower limbs and the same HTO cohort 12 months post-surgery. It is hypothesised that higher medial condyle tibiofemoral joint contact forces would be observed in the KOA cohort pre-HTO compared to the 12 months post-HTO group, whereas the magnitude and position of the knee forces would become normalised post-surgery.

The novel findings from this study demonstrate that at 12-months post-HTO, patients had significantly lower FP and SP medial compartment tibiofemoral contact forces compared to the controls. This reduction is observed 1-year post-surgery as well as a significant increase in walking speed, typically known to increase joint loading (Lenton et al., 2018).

This study determined that at 12-months post-HTO, point of force application was normalised to that of the control cohort except for two metrics (SP medial and lateral compartment loading locations which were lateralised 12 months post-HTO). Finally, this study indicates that the tibiofemoral contact area is normalised for all contact area metrics at 12 months post-HTO in a medial KOA cohort when compared to controls. These findings support the clinical aim of HTO surgery which is to unload the medial compartment of the tibiofemoral joint (Black et al., 2018), to prevent further degeneration of the joint whilst also increasing gait speed.

Previous research by Bhatnagar and Jenkyn (2010) assessed internal knee joint kinetics pre- vs post-HTO. Bhatnagar and Jenkyn found that medial knee compartment load at first peak EKAM pre-HTO was 1.72 %BW which was significantly reduced to 1.16 %BW 12-months post-HTO. Albeit a different metric being assessed, the current paper is in agreement with the findings of Bhatnagar and Jenkyn that medial compartment contact force magnitude is reduced post-HTO when compared with pre-surgery.

This is the first time the COMAK framework has been used to estimate tibiofemoral contact forces and pressures in a cohort of individuals pre- and 12 months post-HTO. Before surgery the medial contact forces were lower in the patients than that of the control cohort at FP

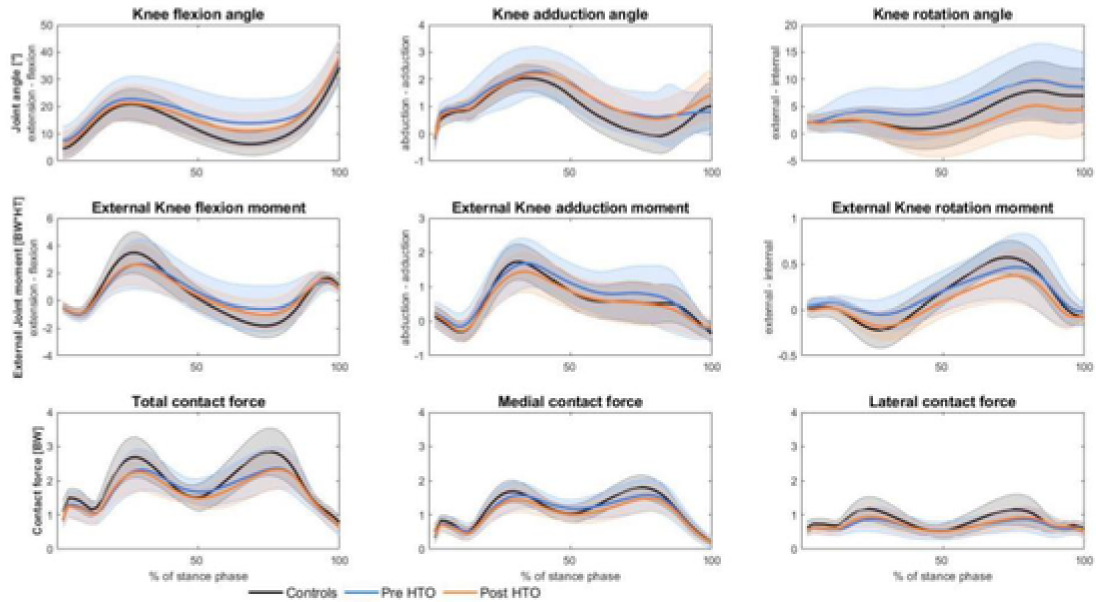

Fig. 1. Pre-and post-HTO COMAK: Knee kinematics, external moments, & contact forces.

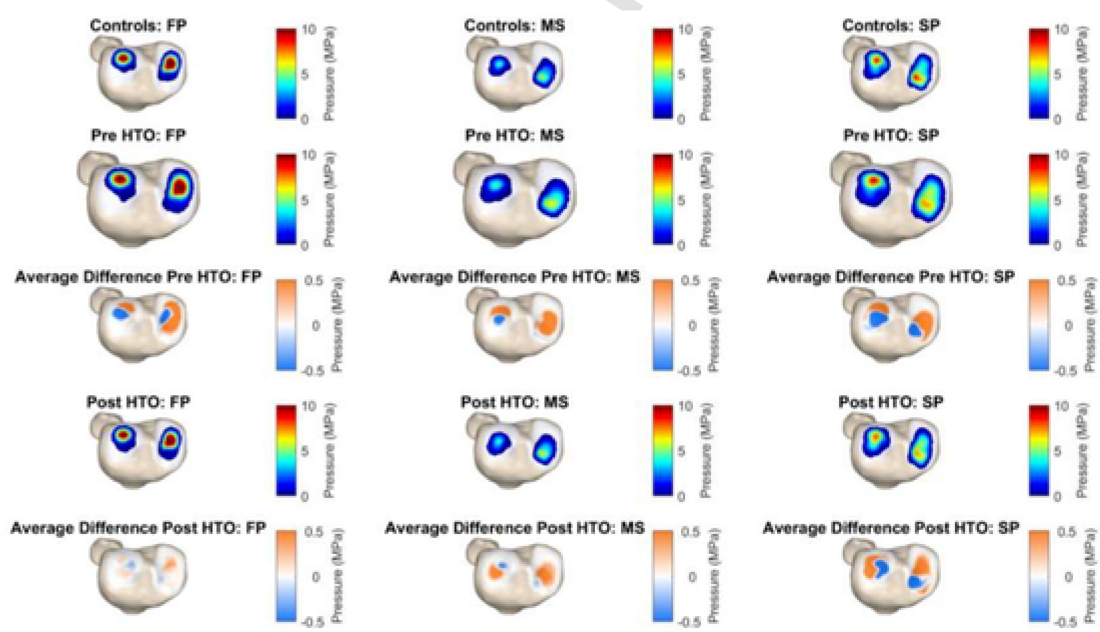

Fig. 2. Pre-and post-HTO contact pressure distribution on the tibia.
Average contact pressure patterns at first peak, midstance and second peak for the control group and the patients pre- and post-HTO. Furthermore, the average difference between the pressure pattern in patients and the healthy control pressure pattern is shown. Orange indicates more loading in the patient on that specific location, blue indicates decreased loading compared to the controls.

and SP. The most significant findings from this research are that HTO reduces medial knee contact force with no negative consequences on lateral compartment contact force. In addition to this, surgery significantly lateralised the centre of pressure for the total knee at first peak, second peak and at midstance. This is the first time the COMAK pipeline has been applied to individuals pre- and 12 months post-HTO. The results therefore go beyond those reported previously: (1) reporting external knee joint moments as a surrogate for medial compartment joint loading (Whatling et al., 2019), (2) systematically simulating a varus knee (van Rossom et al., 2019), and (3) estimating tibiofemoral joint contact forces for a KOA group (Richards et al., 2018). The novel findings support HTO as a surgical intervention to offload the medial compartment of the tibiofemoral joint and should be explored further with patient-specific modelling.

Although the results of the force values are higher in the control cohort compared to the pathological cohort at FP and SP, the pressure (e.g., load interacting with the surface) is higher in pre-surgery relative to controls (except max lateral pressures at some timepoints). Surgery seems to "restore" the pressure to be like the control cohort. This is an important finding from the current paper which has implications for pathology mechanisms. This finding indicates that lower limb varus deformity increases pressure relative to controls. This indicates that pressure causes stress in the joint tissue, which in turn determines the amount of strain, and whether the material will fracture. This is rele-

Table 3
Point of application of the contact forces.

	Controls	Pre-HTO	Post-HTO	Controls vs pre-HTO	Controls vs post-HTO	Pre vs Post HTO
	Mean (std)	Mean (std)	Mean (std)	*P* value	*P* value	*P* value
First peak						
Total knee						
Ant (+) / Post (−)	−2.46 (2.08)	−2.10 (2.93)	−2.56 (2.2)	0.799	0.728	0.186
Lat (+) / Med (−)	−2.27 (3.27)	−5.20 (4.4)	−2.44 (3.43)	**0.006****	0.845	< **0.001**††
Medial knee						
Ant (+) / Post (−)	−1.01 (2.25)	−0.38 (3.24)	−1.18 (2.12)	0.492	0.811	0.075
Lat (+) / Med (−)	−17.19 (1.29)	−18.11 (1.5)	−17.44 (1.53)	**0.015***	0.493	**0.016**†
Lateral knee						
Ant (+) / Post (−)	−4.67 (1.83)	−5.11 (2.46)	−4.65 (2.27)	0.170	0.847	0.223
Lat (+) / Med (−)	20.59 (2.34)	19.63 (2.12)	20.89 (2.08)	0.107	0.614	< **0.001****
Midstance						
Total knee						
Ant (+) / Post (−)	4.66 (1.86)	4.12 (2.24)	4.64 (1.96)	0.325	0.974	0.278
Lat (+) / Med (−)	−5.84 (4.72)	−7.7 (7.04)	−5.47 (5.84)	0.246	0.791	**0.046***
Medial knee						
Ant (+) / Post (−)	6.63 (2.77)	6.41 (2.63)	6.34 (2.74)	0.755	0.688	0.899
Lat (+) / Med (−)	−16.33 (1.53)	−17.10 (2.09)	−16.65 (1.82)	**0.036**	0.453	0.188
Lateral knee						
Ant (+) / Post (−)	−0.42 (1.55)	−2.16 (2.17)	0.32 (1.74)	**0.001****	0.090	< **0.001**††
Lat (+) / Med (−)	20.34 (2.43)	18.91 (3.49)	20.88 (3.22)	0.077	0.471	**0.003****
Second peak						
Total knee						
Ant (+) / Post (−)	5.70 (1.64)	4.45 (3.31)	4.94 (3.22)	0.349	0.717	0.436
Lat (+) / Med (−)	−1.39 (2.9)	−2.69 (5.07)	−0.68 (3.24)	0.065	0.417	**0.007**††
Medial knee						
Ant (+) / Post (−)	9.56 (1.83)	8.13 (3.34)	8.26 (3.68)	0.180	0.201	0.847
Lat (+) / Med (−)	−14.59 (1.31)	−15.85 (1.74)	−15.46 (1.27)	**0.003****	**0.013***	0.140
Lateral knee						
Ant (+) / Post (−)	−0.89 (2.43)	−2.59 (3.34)	−0.39 (2.73)	**0.030**†	0.466	< **0.001****
Lat (+) / Med (−)	19.74 (2.84)	19.49 (3.34)	21.55 (2.9)	0.761	**0.020***	< **0.001****

Point of application (mm) of the total knee, medial and lateral contact force expressed in the tibial reference frame at FP, SP and MS are shown. Ant = anterior, Post = posterior, Lat = lateral, Med = medial. Significant difference ($p < 0.01$) indicated by ** where parametric or †† where non-parametric tests used.

Table 4
Contact area.

	Controls	Pre-HTO	Post-HTO	Controls vs pre-HTO	Controls vs post-HTO	Pre vs Post HTO
	Mean (std)	Mean (std)	Mean (std)	*P* value	*P* value	*P* value
First peak						
Total	352.35 (39)	362.26 (49.94)	362.97 (49.26)	0.357	0.453	0.923
Medial	206.8 (22.63)	223.16 (29.08)	213.55 (32.49)	**0.012**†	0.472	**0.030***
Lateral	145.55 (20.95)	139.1 (29.5)	149.43 (21.45)	0.638	0.242	**0.030**†
Midstance						
Total	262.26 (34.1)	285.17 (55.54)	282.31 (46.27)	0.111	0.080	0.718
Medial	167.14 (24.46)	188.99 (36.33)	180.35 (29.87)	**0.010***	0.108	0.091
Lateral	95.12 (19.62)	96.18 (35.19)	101.96 (27.28)	0.847	0.111	0.345
Second peak						
Total	399.23 (90.44)	381.58 (90.11)	387.36 (90.55)	0.444	0.620	0.702
Medial	219.07 (41.52)	222.87 (49.9)	217.24 (46.55)	0.755	0.763	0.515
Lateral	180.16 (52.6)	158.71 (47.48)	170.12 (46.05)	0.108	0.442	0.132

Contact area (mm^2) in the total knee, medial and lateral compartment at first peak, second peak and midstance are shown. Significant difference ($p < 0.01$) indicated by ** where parametric or †† where non-parametric tests used.

vant for the relationship between joint loading and cartilage damage associated with mKOA.

Some limitations need to be considered when interpreting the results of this study. The COMAK model's calculation of abduction/adduction was not validated in this current study. Additionally, this current study did not account for possible patient-specific long axis rotation alterations. In terms of the methodology, the model comprises a generic knee model, with a uniformly distributed articular cartilage thickness. In medial compartment KOA, cartilage thickness can be reduced and variable across the articulating surfaces (Agnesi et al., 2008) which may affect contact loading parameters. Thus, for the current study, the effect of variations in articular cartilage thickness in the KOA cohort on the calculated contact pressure distribution are neglected. This simplification might result in differences in terms of contact pressures and contact areas. Therefore, the findings from the current study may underestimate the contact forces and future work should aim to incorporate a more patient-specific approach to defining knee cartilage and bone geometry. For the Lenhart model, the tibiofemoral adduction is defined by the geometry of the contact surfaces and so it was not feasible to manually alter this. Therefore, it was decided to adjust the tibial and foot segments, i.e., ankle joint. The ankle is translated to a new location based on the mTFA angle to effectively introduce varus/valgus and corrected for foot alignment, so it lands flat on the floor.

Additionally, the optimisation algorithm used in the current study did not account for subject-specific muscle contractions. This would require an EMG-driven modelling approach. Future work should be undertaken to establish the influence of co-contraction on the predictive manner of COMAK as recent work has shown HTO patients have a different neuromuscular pattern compared to a non-pathological cohort (Ghazwan et al., 2022).

The control cohort involved in this study were not aged matched. Although the mean age of the control cohort is lower than the patient cohort, this study shows that following HTO, biomechanical measures of knee loading are improved to values like the younger control group, adding weight to the proposed merits of HTO surgery.

Additionally, the control cohort in this study were not BMI matched to the pathological group; the HTO cohort had significantly larger BMI compared to both pre- and post-HTO. Body mass influences joint moment magnitudes. This work attempted to factor this potential confounding variable by presenting contact forces as a multiple of body weight.

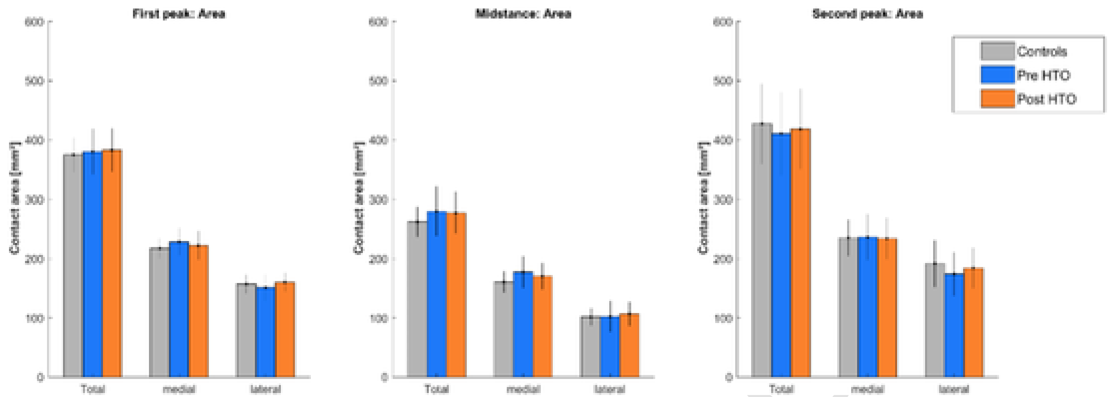

Fig. 3. Pre- and post-HTO contact area.

Finally, gait speed was not accounted for statistically. The patient cohort had medial KOA and therefore it was considered inappropriate to correct for gait speed when assessing joint loading changes, as discussed by Wilson (2012), who suggested that when accounting for gait speed in statistical analysis for KOA patients, critical assumptions of the statistical model are violated making the methods inappropriate and the results misleading.

5. Conclusions

This study has applied a generic musculoskeletal model that incorporates a detailed knee model to understand tibiofemoral contact loading in medial knee osteoarthritis and estimated the effects of HTO surgery.

HTO successfully re-aligned the lower limbs, increased gait speed, reduced medial compartment contact force and mean pressure during the first half of stance, altered the application of force during stance, as well as reducing the medial compartment contact are at FP.

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Ethical approval

Approval for this work was granted by the Wales Research Ethics Committee 3 (10/MRE09/28) and Cardiff and Vale University Health Board.

Declaration of Competing Interest

The authors declare that no conflicts of interest exist.

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