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Effects of an intensive weight loss program on knee joint loading in obese adults with knee osteoarthritis

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SUMMARY

Objective: To determine the effect of an intensive weight loss program on knee joint loads during walking in obese patients with knee osteoarthritis (OA).
Methods: Participants included 157 obese knee OA patients that underwent a 16-week dietary intervention. Three-dimensional gait analyses were performed before and after the intervention at the participants' freely chosen walking speed. Knee joint compression forces, axial impulses, knee flexion angle and frontal and sagittal plane knee moments were calculated to determine the biomechanical effects of the weight loss.
Results: 157 subjects (89% of the initial cohort) completed the 16-week intervention. The average weight loss of 13.7 kg ($P < .0001$) corresponded to 13.5% of the baseline body weight. The weight loss resulted in a 7% reduction in knee joint loading, a 13% lower axial impulse, and a 12% reduction in the internal knee abductor moment (KAM). There were no clear effects on sagittal plane knee moments or peak knee flexion angle. Linear regression analyses adjusted for changes in walking speed showed that for every 1 kg in weight loss, the peak knee load was reduced by 2.2 kg. Thus, every kilo reduction in body weight was related to more than twice the reduction in peak knee force at a given walking speed.
Conclusion: Weight loss is an excellent short-term investment in terms of joint loading for patients with combined obesity and knee OA. The effects of sustained weight loss on disease progression and symptoms in relation to biomechanical factors remain to be shown.

Introduction

Osteoarthritis (OA) is a common disease of multifactorial origin. Facing extensive and increasing economic and social burdens due to the complications of OA, limiting disease progression is an important public health strategy[1]. It is well established that obesity is strongly linked to knee OA and is considered a risk factor for both incidence and progression[2]. Accordingly, weight loss is advocated as the treatment of choice for overweight and obese knee OA patients[3], as it yields clinically significant reductions in pain and improvements in function[4].

The pathway by which obesity is thought to cause knee OA is through increased joint loading[5–8]. The association between obesity and joint load during walking is intuitive, and it is generally accepted that knee OA is biomechanically driven[7,9]. In fact, the peak internal knee abductor moment (KAM) reflects the load distribution on the tibial plateau[10] and its magnitude is a strong predictor of presence[11] and rate of progression[6] of medial knee OA. Therefore identifying mechanisms to reduce the internal KAM is imperative in preventing knee OA progression. Accordingly, weight loss would be expected to reduce medial joint loads, and possibly delay disease progression for patients with medial knee OA.

While the peak forces in the knee represent maximum compressive loadings, impulses provide a measure of compressive loading over time and across the stance phase. Such total knee forces and impulses together with the sagittal and frontal plane joint moments may represent important changes in the gait pattern. Yet, to our knowledge only one study has targeted the effects of weight loss on total dynamic joint loadings during walking in obese knee OA patients[12]. That study showed a four-fold decrease in joint loads for each lost unit of body weight. However, the weight loss was modest (2.6%)[12], and although significant reductions in joint loads during walking were reported, a meta-analysis showed that at least 10% weight reduction is necessary if clinically significant effects on pain

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and disability should be expected[4]. Further, consensus among obesity specialists is that at least 10% weight loss is necessary to reduce other risk factors associated with being overweight or obese, such as cardiovascular diseases and type 2 diabetes[13,14]. The biomechanical consequences of such significant weight loss (>10%) in obese knee OA patients are not known.

The purpose of this study was to evaluate the effect of an intensive weight loss program aimed at a >10% weight loss on knee joint loadings during walking in obese knee OA patients. We hypothesised that the weight loss would induce a significant reduction in joint loading.

Patients and methods

Data from 177 knee OA patients included in the dietary intervention study; "The Influence of Weight Loss or Exercise on Cartilage in Obese Knee OA Patients (The CAROT Trial)"[15] were used in the current study. Eligibility criteria for the patients were: body mass index (BMI) > 30 kg/m^2; age ≥ 50 years, primary knee OA diagnosed according to the American College of Rheumatology criteria[16], with clinical symptoms and radiographically or arthroscopically verified OA in one or both knees. The CAROT-study was approved by the local ethical committee [ref.: H-B-2007-088]. Patients not able to walk independently without a walking aid were excluded from the present study. All measurements were performed before and after the weight loss intervention.

Weight loss program

As part of the CAROT-study the patients followed a supervised dietary program in a 16-week intensive treatment period, consisting of a hypo-energetic diet of normal food plus meal replacements (The Cambridge Diet, the Cambridge Health and Weight plan UK) and nutritional education. The nutritional instructions and behavioural therapy were provided by an experienced dietician at weekly sessions (1.5 h/week) throughout the 16 weeks to reinforce the patients' decision about weight reduction and to encourage a high degree of compliance. The dietary program met all recommendations for daily intake of vitamins and minerals. The goal of the dietary program was to reduce body weight by at least 10% and is described in detail elsewhere[15].

Gait analysis

Kinematic data were acquired using a three-dimensional (3D) motion analysis system (Vicon MX, Vicon Motion Systems, Oxford, UK) with six cameras (MX-F20, Vicon Motion Systems, Oxford, UK) operating at 100 Hz. Two force platforms (AMTI OR 6-5-1000, Watertown, MA, USA) embedded in the laboratory floor captured ground reaction forces at 1500 Hz synchronized with the kinematic data. The 3D orientation of seven body segments of interest (pelvis; left and right thighs; left and right shanks; both feet) was obtained by tracking trajectories according to a common commercially available kinematic model (Plug-In-Gait, Vicon Peak, Oxford, UK), with markers placed bilaterally on the anterior and posterior iliac spines, lateral aspect of the thighs, lateral femoral epicondyles, lateral aspects of the shanks, lateral malleoli, calcanea, and second metatarsal heads. Markers were placed directly on the skin and patients walked barefooted during all trials.

Initially, markers were placed at anatomical landmarks and a static calibration trial was captured. Subsequently, the patients walked the 10 m walkway freely until a stable and comfortable walking speed was obtained. The mean of six practice trials at this speed determined the preferred walking speed. A photocell system registered the walking speed with a digital display providing the subjects with immediate visual feedback. The starting point was adjusted for each subject to ensure a clean foot strike on either of the two force platforms. Once walking speed and starting points were determined, a series consisting of six acceptable trials, within ±0.1 km/h of target speed, with successful force platform hits without observable targeting, were performed.

The patients had their affected knee analyzed; if both knees were affected, the most symptomatic knee was chosen for analysis based on patient report. The analyses focused on the stance phase of the gait cycle which was defined as from heel-strike (HS) to toe-off (TO) (when the vertical ground reaction force exceeded 5 N). Kinematic marker coordinate data were filtered using Woltring's generalized cross-validation quintic smoothing spline with a predicted mean square error of 15 mm. 3D joint moments, reaction forces, and knee flexion angles were calculated using an inverse dynamics approach using the Plug-In-Gait model (Vicon Motion Systems, Oxford, UK). The peak values of internal stance phase joint moments, joint reaction forces and joint angles were used for the modelling of joint compressive forces (see later). Kinetic peak values extracted for statistical analyses are illustrated in Fig. 1. All trials were analyzed individually (i.e., no within subject averaging).

Knee joint compressive force and axial impulse

To assess the knee joint compressive forces we applied a statically determinate muscle model previously published[10]. The overall knee compression force was estimated as the vector sum of (1) the intersegmental reaction forces resolved along the long axis of the tibia, (2) the compression components of the active muscle group forces and (3) the axial component of the cruciate ligament forces. The included muscles were the hamstrings, gastrocnemius and quadriceps muscles. The hamstring and gastrocnemius complex constituted a flexor muscle group active when the net internal sagittal knee joint moment favoured the flexors, and the quadriceps muscle represented an extensor muscle group active when the net moment favoured extensors. The muscle forces were estimated by dividing the net sagittal plane joint moments with the muscle moment arms. The moment arms were derived from a third-order polynomial that estimates the moment arms based on the sagittal plane knee joint angle[17]. The axial cruciate ligament forces were estimated under the assumption that the cruciates only resist antero-posterior shear forces calculated as the sum of the antero-posterior knee joint reaction force and the antero-posterior component of the muscle forces acting over the knee. The medio-lateral position of the tibio-femoral contact point is fixed at 25% of the knee joint diameter from the knee joint center, whereas the anterior–posterior contact point changes with knee joint flexion[10]. In the original model[10] the knee joint diameter was fixed at 80 mm for all subjects. In the present study the knee joint diameter was obtained from each subject by means of a calliper as a part of the anthropometric measurements required for the gait analysis. The knee joint diameter was used for calculation of the position of the tibio-femoral contact point. Thus, the moment arm of the lateral soft-tissue that counteracts the external adduction moment was scaled by the knee joint diameter. Peak compressive forces (during the last 90% of the stance due to the artificial nature of the first peak, see Fig. 1) and axial impulse (i.e., area under the entire time-peak compressive force curve) were extracted [Fig. 1(C)]. The peak flexion knee angle was determined as the peak value during the first half of stance. The calculations used for the model were done in custom written code in Matlab (MathWorks Inc., MA, USA).

Pain scoring

Average knee pain (in the target knee) in everyday life was assessed by a 100 mm visual analogue scale (VAS) with the

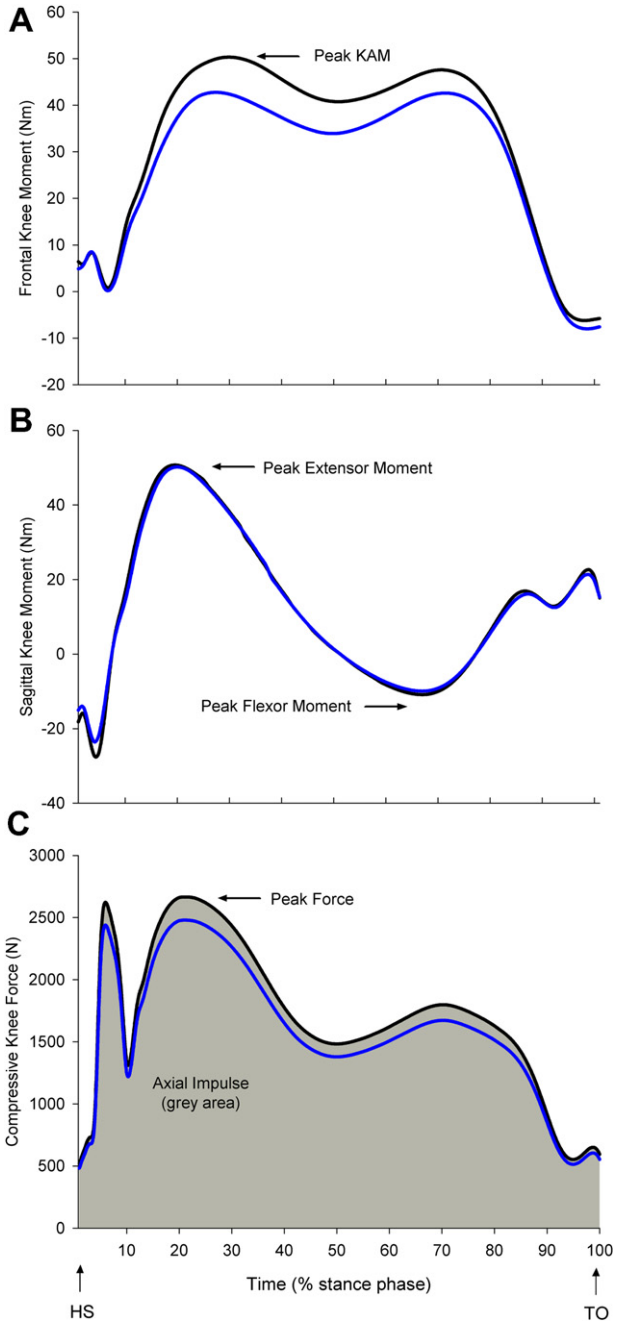

Fig. 1. Ensemble averages showing the time-course pattern of the (A) internal sagittal plane knee joint moment, (B) internal KAM, and (C) compressive knee joint force. Peak moments and forces used in the statistical analyses are shown with text-arrows. Please note that the temporal occurrence of the peak values may not match those indicated on this figure. On Fig. 1C the initial peak is considered artificial and is ignored during the data analysis. On 1C, the axial impulse (i.e., area under the curve) is shown as shaded. HS = heel-strike and TO = toe-off.

extremes anchored in 0 = "no pain" and 100 = "worst imaginable pain"[18].

Radiographic evaluation

Standard semi-flexed standing radiograph was taken (Philips Optimus). A trained musculoskeletal radiologist performed the Kellgren–Lawrence (K/L) score in all standing radiographs as originally described by K/L[19]. Using this method the knee joints were categorized into five grades from 0 to 4, assessing the stage of knee OA.

Alignment

From the static anatomical landmark calibration trials for each subject the knee joint mechanical axis alignment was calculated using the Plug-In-Gait model (Vicon Motion Systems, Oxford, UK). This was defined as the angle in the frontal plane between the lines connecting ankle, knee and hip joints. This angle correlates significantly with the mechanical axis alignment measured from standing full limb radiographs, yet without radiation exposure[20]. A knee was defined as a varus when alignment was $>0^\circ$ and valgus when $<0^\circ$.

Statistical analysis

Data are presented as mean ± standard error (SE) unless otherwise stated. The primary outcome was changes in peak knee joint compression force, while changes in axial impulse, knee angle and the frontal and sagittal plane peak moments were exploratory variables. All trials were used in the statistical analyses, i.e., no within subject averaging. A mixed model with random effects for subject (random intercept) was applied, using the MIXED procedure of the SAS® system (version 9.1.3; SAS Institute, Cary, NC, USA). The analyses focused on the effects of changes in body weight following the dietary intervention on changes in knee joint forces and moments. The crude analyses were repeated including age, gender, knee alignment axis and changes in both walking speed and pain as covariates. Also height was included as a covariate in the analyses of joint moments. For each variable the mixed model produced a best-fit linear regression equation from which the slopes (beta-coefficients) of the linear fits were extracted. To assess if the slopes were significantly different from zero a T score (beta coefficient divided by SE) was computed and a Student's two-tailed *t*-test was applied. Statistical significance was accepted at $P < 0.05$.

Results

157 (89% of the initial cohort) patients had their gait analysis repeated at follow-up. Student's *t*-test revealed no significant differences between the 20 dropouts and the 157 completing patients in baseline characteristics ($P > 0.16$). Baseline characteristics of the participants are given in Table I. The peak knee loadings corresponded to 2.7 times body weight at baseline (Table I).

The diet intervention resulted in an average body weight loss of 13.7 kg [95% confidence interval (CI) −12.9 to −14.4 kg, $P < .0001$] corresponding to 13.5% of the baseline average body weight (Table II). Concomitantly, BMI was reduced by 5.1 kg/m^2 (95% CI −5.3 to −4.8 kg/m^2, $P < 0.0001$) equal to 13.7% from 36.9 at baseline to 31.9 at follow-up. Self-selected walking speed was increased by 4% (mean difference: 0.04 m/s; 95% CI 0.02–0.07 m/s; $P < .0001$), while pain decreased by 30% (mean difference: −13 mm; 95% CI −10 to −16 mm; $P = < 0.0001$). Knee joint flexion angle was slightly but significantly reduced from 18.9° before to 18.5° after weight loss ($P = 0.011$).

A summary of the changes in joint forces and moments are given in Table II. Following the weight loss, the average reduction in peak knee compression force was 189.9 N (95% CI 98.7–281.2 N; $P = < .0001$) corresponding to 7.0% of the baseline value. When the covariates age, gender, knee alignment axis and changes in both walking speed and pain were included the mean reduction was slightly lower but still significant (mean adjusted reduction 157.1 N; 95% CI: 41.7–272.4 N; $P = 0.008$). The unadjusted axial impulse was significantly reduced following weight loss by an average of 148.2 Ns (95% CI 206.9–93.3 Ns; $P = < .0001$) corresponding to 13% of the baseline value. Axial impulse change was not altered by

Table I
Baseline characteristics of obese knee OA patients

	Baseline ($n = 157$)
Gender	
Females, no. (%)	129 (82%)
Males, no. (%)	28 (18%)
Age, years	62.7 ± 6.4 (61.7–63.7)
Height, m	1.66 ± 0.08 (1.65–1.68)
Body mass, kg	101.5 ± 14.1 (99.2–103.7)
BMI, kg/m^2	36.9 ± 4.3 (36.2–37.6)
Self selected walking speed, m/s	1.12 ± 0.19 (1.09–1.15)
Alignment (degrees, positive is varus)	5.8 ± 4.8 (5.0–6.5)
Pain (0–100 score, lower is less pain)	42.8 ± 20.6 (39.6–46.0)
Tibio-femoral K/L score, no. (%)	
Grade 0	0 (0%)
Grade 1	25 (16%)
Grade 2	59 (38%)
Grade 3	47 (30%)
Grade 4	22 (17%)
Lateral/medial/patellofemoral involvement, no. (%)	140 (89%)/150 (96%)/148 (94%)

Except where indicated otherwise, results are given in mean ± standard deviation (95% CI).

including age, gender, knee alignment axis and changes in both walking speed and pain in the model (Table II).

In the adjusted model the best-fit linear regression analyses showed that for every 9.8 N loss in body weight, the peak knee joint force was reduced by 21.7 N (95% CI 2.4–41.0 N; $P = 0.029$; Table III). When we explored the differences between the crude and adjusted analyses the only covariate that contributed significantly to the regression was change in walking speed (main effect: $P = 0.0003$) showing that for each 0.1 m/s increase in walking speed the peak joint load increased by ~44 N, which counteracts the effects of weight loss. Thus, the adjusted model shows that at a given walking speed every kilo of weight loss leads to more than twice the reduction in peak knee load (21.7 N divided by 9.82 N = 2.2 kg). The axial impulse was reduced by 13.9 Ns/kilo weight loss (95% CI 4.4–24.4 Ns/kg; $P = 0.005$) and the inclusion of covariates did not alter this association (Fig. 2).

The sagittal plane internal knee joint moments were not changed following weight loss ($P \geq 0.20$; Table II). Accordingly, unadjusted beta-coefficients for the extensor and flexor moments were not significantly different from zero: 0.18 Nm/kg (SE 0.35 Nm/kg; $P = 0.60$) and −0.04 Nm/kg (SE 0.30 Nm/kg; $P = 0.89$) extensor and flexor moments respectively. Including covariates in the model did not change these results. However, the internal KAM was lowered significantly by 6.3 Nm (95% CI −7.7 to −4.9; $P < .0001$) corresponding to 12% of the baseline value and this change magnitude was unchanged when we included covariates in the analysis (Table II). The adjusted beta coefficient for the change in internal KAM demonstrated a significant association with changes in body weight, showing a reduction of 0.6 Nm/kilo body weight lost (SE 0.15 Nm/kg; $P = 0.0002$, Table III).

Table II
Change after weight loss

Variable	Baseline	Change	% Change	P
Body mass, (kg)*	101.5 (1.1)	−13.6 (0.2)	−13.2%	<0.0001
BMI (kg/m2)*	36.9 (0.4)	−5.1 (0.1)	−13.8%	<0.0001
Pain (0–100 score, lower is better)*	42.8 (1.6)	−13 (1.6)	−30.4%	<0.0001
Self selected walking speed (m/s)*	1.12 (0.01)	0.05 (0.13)	4%	0.017
Peak flexion knee angle (°)	18.9 (0.5)	−0.4 (0.15)	−2%	0.011
Compressive force (N)†				
Adjusted‡	2695.4 (68.5)	−157.1 (58.4)	−6%	0.008
Unadjusted	2654.9 (59.0)	−189.9 (46.2)	−7%	<0.0001
Axial impulse (Ns)†				
Adjusted‡	1149.6 (39.7)	−150.1 (28.8)	−13%	<0.0001
Unadjusted	1118.0 (30.1)	−148.0 (21.5)	−13%	<0.0001
Peak internal extensor moment (Nm)†				
Adjusted‡	46.2 (2.4)	0.42 (2.2)	1%	0.84
Unadjusted	50.2 (2.1)	−1.72 (1.6)	−3%	0.28
Peak internal flexor moment (Nm)†				
Adjusted‡	−20.1 (1.8)	2.60 (2.0)	13%	0.20
Unadjusted	−11.2 (1.6)	−0.53 (1.3)	−5%	0.69
Peak internal abductor moment (Nm)†				
Adjusted‡	53.68 (1.5)	−6.35 (1.0)	−12%	<0.0001
Unadjusted	50.34 (1.6)	−6.30 (0.7)	−13%	<0.0001

Values are mean (SE).
N = Newtons and s = seconds. P-values indicate whether changes are significantly different from zero. Level of significance $P < 0.05$.
* Based on student's t-test.
† Based on analysis of variance (ANOVA).
‡ Adjusted for age, gender, knee joint alignment axis and changes in walking speed and pain. Height was included as covariate in the analyses of joint moments also.

Table III
Best linear fit regression slopes (beta-coefficients) for the association between the changes in variables and change in body weight

Variable	Slope	T score	P
Compressive force (N/kg)†			
Adjusted*	21.71	2.20	0.029
Unadjusted	17.13	1.70	0.091
Force impulse (Ns/kg)†			
Adjusted*	13.86	2.86	0.005
Unadjusted	15.83	3.37	<0.001
Peak extensor moment (Nm/kg)†			
Adjusted*	0.31	0.92	0.36
Unadjusted	0.18	0.52	0.60
Peak flexor moment (Nm/kg)†			
Adjusted*	−0.06	−0.19	0.84
Unadjusted	−0.04	−0.13	0.89
Peak internal abductor moment (Nm/kg)†			
Adjusted*	0.55	3.46	<0.001
Unadjusted	0.48	3.16	0.002

N = Newtons and s = seconds. Statistical significance was accepted at $P < 0.05$.
* Adjusted for age, gender, knee joint alignment axis and changes in walking speed and pain. Height was also included as covariate in the analyses of joint moments.
† Level of significance $P < 0.05$.

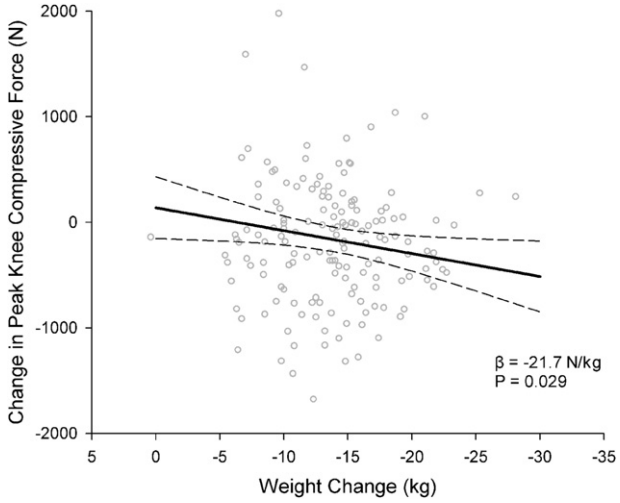

Fig. 2. Scatter plot illustrating the relationship between weight loss and change in knee joint compressive force adjusted for age, gender, alignment and changes in both walking speed and pain. Full line shows the best-fit linear regression with 95% confidence intervals (dotted lines). The slope (β-value) is significantly different from zero ($P = 0.029$).

We calculated the sample size needed to detect the present changes in knee joint loading with a power of $\beta = 80\%$ and $\alpha = 5\%$ and we found that 123 pairs of observations would be sufficient.

Discussion

This study shows that weight loss significantly reduced knee joint loads during walking. However, a concomitant increased self-selected walking speed interfered with this reduction in joint loads. When adjusting for changes in walk speed, every kg of weight loss lowered the peak knee joint load by 2.2 kg. As such, the change in peak compressive force due to weight loss confirmed our hypothesis of a beneficial effect of the weight loss. Furthermore, the axial impulse and KAM were also significantly reduced by 13% and 12% respectively.

Peak compressive loads on the knee joint tend to be 2–4 times body weight[21–23] which is similar to our findings at baseline (2.7BW). Furthermore, our results corroborate previous studies of weight loss and joint loads during walking[12,24]. However, while Messier *et al.* found that for each unit of weight loss the joint loads were reduced by 4 units, in this study a 2:1 relationship was found. There may be several explanations for this. Firstly, our weight loss intervention yielded a 13.5% weight loss compared to 2.6% in[12]. Secondly, in the present study the range of weight loss was 28.5 kg (max: −28.1 kg; min: 0.4 kg) and although the weight loss range was not given by Messier *et al.* [12], a larger range gives better mathematical basis for linear regressions[25,26]. Further, the authors[12] used a different biomechanical model to estimate the joint loads, including ankle and hip kinetic contributions to the knee joint loads, whereas we used a model that only incorporated knee joint mechanics. However, a 2:1 reduction in knee compressive load compared to weight loss seems reasonable in the light of previous studies[21–23].

Overweight knee OA patients have been shown to use a kinetic coordination strategy that reduces the contribution from the knee and increases contributions from the ipsilateral ankle joint[27]. Further, the weight loss intervention brought about an increase in the self-selected walking speed, and joint loads are positively related to walking speed[28]. The overall purpose of this study was to evaluate the biomechanical effects of weight loss. One of these biomechanical variables is walking speed and the follow-up gait analysis was therefore performed with a new self-selected walking speed to detect this effect. In fact, our results show that greater walking speed following weight loss increased the joint loads significantly (~44 N increase/0.1 m/s speed increment) and thus counteracted the reduction in joint loads. Indeed, our results showed that adjusting for walking speed was necessary to reveal the association between change in peak joint load and weight loss.

Our results confirm the previous observation[12] that the peak knee compressive force was reduced more than what can be accounted for by the weight loss alone. In a cross sectional study, BMI was found to have a strong, inverse relationship to knee extensor moment during self-selected walking speed in obese but otherwise healthy subjects[29]. Accordingly, a weight loss would be expected to induce a higher internal extensor moment that is associated with increased joint loads. However, our longitudinal results show that the internal knee extensor moments were unaffected by the weight loss. The inverse relationship between BMI and the knee extensor moment observed in[29] may be kinematically explained by less knee flexion angle reducing the knee extensor moment arm. However, in the present study the weight loss reduced the knee flexion angle during stance marginally (2%). As such, the weight loss did not affect either sagittal knee joint moments or angles substantially which diminishes the likelihood of neuromuscular adaptations to occur in this plane.

Generally knee OA patients walk with reduced internal knee extensor moments to reduce joint loads[30], a gait strategy that is presumably pain driven[31]. In the present study, pain was significantly reduced following the weight loss, supporting previous findings that load reducing interventions attenuate clinical OA symptoms[32,33]. Conversely, pain relief relates to increases in knee joint loading during walking and stair climbing[34–37]. In the present data changes in pain did not affect the change in joint loads, and it is possible that the reduction in joint loads due to weight loss prevail over any increase in joint loads due to the reduced pain.

The axial impulse represents the total or cumulative mechanical load on the knee. We observed a significant 13% reduction in axial impulse after the weight loss independent of change in walking speed. While the clinical relevance of a reduced axial impulse currently is unknown the lowering by 12% of the internal KAM (a proxy for medial compartment loading[10,22]) may have implications especially for medial knee OA patients. The internal KAM has been shown to predict presence of bone marrow lesions in knee OA patients[38] and it is the strongest predictor of medial knee OA progression[6]. Thus the abductor moment is more relevant for patients with dominating medial knee OA and a reduced abductor moment may halt the disease progression in these patients. We included both lateral, patellofemoral, and medial knee OA patients in the present study and 14, 27, and 61 patients demonstrated predominant lateral, patellofemoral, or medial knee OA involvement, respectively. Also, 55 patients demonstrated equally severe OA in two or more compartments. While the internal KAM has been shown particularly important for medial OA, it is unknown how total knee joint loads reductions, as presented in this study, is affected by different compartmental OA involvements. However, due to the skewed distribution of compartmental involvements such analyses would be subject to considerable ambiguity as it would be underpowered in the present material.

It is important to acknowledge the study limitations. Only per protocol analyses were performed since our study aim focused on the biomechanical effect of weight loss on joint loads and not the clinical effect of a weight loss intervention. Thus, our results primarily apply to the understanding of basic mechanics and the effects of weight loss in both medial and lateral knee OA. Also, in our model we did not incorporate co-contractions and thereby we may have underestimated knee joint forces. There are other important limitations in the knee joint model used in the present study. In particular, the validity of motion analysis on overweight/obese subjects is limited by the difficulties in marker placements and acquirement of anthropometry (e.g., knee joint diameters). However, *in vivo* measurement of joint loadings are not feasible in this population, and joint loadings directly measured are in close agreement with measurements obtained from non-invasive motion capture analysis[39]. Also, using motion capture for estimates of knee joint alignment has not been validated in obese subjects, but in the present study we did not have access to full limb radiographs. Furthermore, the model is for most parts generic, based on estimates from the literature (e.g., muscle moment arms), and subject specific measurement, such as from magnetic resonance imaging (MRI), may yield other results. The gait analyses were performed barefooted which may limit real-world applicability of the results. However, barefooted measurements were preferred in this study because shod measurements may be flawed by changes in footwear across time and seasons.

Because we included 157 pairs in the present study it is concluded that the results from the present study are adequately powered. Weight loss is an evidence-based treatment for knee OA[3,4,40], and a 5.1 kg reduction over a 10-year period decreases the likelihood of developing knee OA by 50%[41]. Therefore, our results suggest that reducing joint loads with an intensive weight loss may

slow disease progression. In the future, it is of great importance to investigate the effect of a substantial weight loss on long-term knee OA progression.

In conclusion, we showed that, after controlling for walking speed, each unit of weight loss was related to a reduction in knee joint loads by a factor of 2.2. We also found that the cumulative load (axial impulse) was significantly decreased by 13% along with a similar reduction magnitude in internal KAM. The decreased axial load was offset by changes in walking speed, and the benefits of weight loss may be more related to changes in cumulative load (axial impulse) or internal KAMs. Weight loss is an excellent short-term investment in terms of biomechanical estimates of joint loadings for patients with combined obesity and knee OA.

Contributions

All authors have made substantial contributions to all three of sections (1), (2) and (3):

(1) The conception and design of the study, or acquisition of data, or analysis and interpretation of data (2) drafting the article or revising it critically for important intellectual content (3) final approval of the version to be submitted.

Conflict of interest

Aaboe, J., Bliddal H., and Henriksen, M. received travel grants to attend scientific meetings from Cambridge Weight Plan.

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