Walking Velocity and Step Length Adjustments Affect Knee Joint Contact Forces in Healthy Weight and Obese Adults

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ABSTRACT: Knee osteoarthritis is a major public health problem and adults with obesity are particularly at risk. One approach to alleviating this problem is to reduce the mechanical load at the joint during daily activity. Adjusting temporospatial parameters of walking could mitigate cumulative knee joint mechanical loads. The purpose of this study was to determine how adjustments to velocity and step length affects knee joint loading in healthy weight adults and adults with obesity. We collected three-dimensional gait analysis data on 10 adults with a normal body mass index and 10 adults with obesity during over ground walking in nine different conditions. In addition to preferred velocity and step length, we also conducted combinations of 15% increased and decreased velocity and step length. Peak tibiofemoral joint impulse and knee adduction angular impulse were reduced in the decreased step length conditions in both healthy weight adults (main effect) and those with obesity (interaction effect). Peak knee joint adduction moment was also reduced with decreased step length, and with decreased velocity in both groups. We conclude from these results that adopting shorter step lengths during daily activity and when walking for exercise can reduce mechanical stimuli associated with articular cartilage degenerative processes in adults with and without obesity. Thus, walking with reduced step length may benefit adults at risk for disability due to knee osteoarthritis. Clinical Significance: Adopting a shorter step length during daily walking activity may reduce knee joint loading and thus benefit those at risk for knee cartilage degeneration. © 2018 Orthopaedic Research Society. Published by Wiley Periodicals, Inc. J Orthop Res 36:2679–2686, 2018.

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Knee osteoarthritis (OA) is a major public health problem for which there is currently no cure, and adults with obesity are particularly at risk. Approximately 14 million adults in the United States suffer from knee osteoarthritis, half of whom are younger than 45 years old.[1] Thus, the prevalence of knee OA is expected to increase rapidly as our youth with obesity mature into adults with obesity.[1] Recently, attention has begun to focus on the pre-disease stage with the aim of preventing or delaying disease onset.[2] Development and progression of knee OA have been associated with obesity and include both biological and mechanical components.[3] In particular, higher knee adduction impulse has been associated with greater medial tibial cartilage loss over a 1 year period.[4] Impulse variables provide an indication of cumulative load by accounting for load during the whole of the stance phase (rather than a single peak value at one point in time).[5] While weight loss in obese patients is an important step to reducing potentially detrimental mechanical loads,[6] a complementary approach may be to reduce knee joint loads during daily activity through gait modification.[7]

Adjusting knee biomechanics during daily activities, particularly walking, has the potential to reduce cumulative load at the joint. Adults with higher BMI may respond differently to walking cues than those of healthy weight.[8] Therefore, the effectiveness of walking adjustments on knee joint loads in both healthy weight and obese individuals should be established. Other methods to reduce cumulative load per step and other knee variables have been proposed in the literature, including uphill treadmill walking, valgus knee braces, laterally wedged insoles, increased trunk lean, and increased toe out.[7,9,10] However, these approaches require either a device or equipment or large changes to the cosmetic appearance of walking, which may reduce compliance.

Walking is a common daily physical activity during which the knee experiences forces of up to four times body weight with each step.[6] The distribution of this force across the load bearing surfaces of the tibiofemoral joint is reported to have a relationship with knee joint OA. For example, a larger external peak knee adduction moment and a larger knee adduction impulse in adults with OA compared to controls are found in symptomatic individuals[11] and are predictive of disease progression.[4,12–14] However, these biomechanical variables do not necessarily reflect the magnitude of the load directly applied to internal knee structures[15] because knee joint forces during gait are dependent on the combined effects of external and internal muscle forces.[16] Musculoskeletal models can be used to account for both internal and external contributions to knee joint contact forces. Based on such models, temporospatial gait characteristics, for example, step length and gait velocity, have been shown to significantly affect knee joint contact forces.[7,17] As such, modification of these gait characteristics may be relevant to knee OA prevention and treatment efforts.

Previous studies have explored the effects of modifying either walking velocity or step length on knee biomechanics. Taking 15% shorter steps while maintaining velocity during treadmill walking in healthy weight or obese young women decreased knee

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adduction impulse, but not peak knee adduction moment.[18] Sagittal plane changes should also be considered, as a combined reduction in both frontal and sagittal plane knee joint moments may provide a greater reduction in medial compartment load.[19] Decreasing velocity in young adults who were of healthy weight or obese led to decreased peak knee flexion moment in both treadmill[20] and over ground walking.[21] However, the potential for a greater effect of combining velocity and step length changes on knee internal joint loads during walking has not been explored.

Increased joint contact force is a widely regarded knee osteoarthritis etiological risk factor.[22,23] As an example, the incidence and prevalence of knee osteoarthritis is greater in the intact knee of individuals with unilateral amputations relative to the residual leg and non-amputees. Cross sectional data demonstrate increased total TFJ peak contact force[24] and medial tibiofemoral joint peak force and impulse in the intact limb of service members with limb loss compared with a control group of similar age and mass.[25] Though limited prospective evidence exists, it is notable that these mechanical differences were observed prior to the onset of radiographic osteoarthritis in these previous studies. The potential for reduced joint contact forces to facilitate functional articular cartilage repair at the knee has been illustrated by the success of joint distraction surgery. Several clinical studies have demonstrated improvements in cartilage thickness following a period of reduced loading achieved by surgical distraction of the joint.[26] If reduced joint loading provides an environment where cartilage repair can occur, then conservative treatments that reduce joint contact forces would be an attractive option. Interventions to reduce joint contact force prior to the development of osteoarthritis may be desirable, particularly among people with additional risk factors for OA such as obesity. Simple modifications to the velocity and step length during walking may reduce the magnitude of biomechanical risk factors for knee OA. However, the potential for detrimental increases in axial loading at other joints, particularly the hip is a concern with gait modification.[27] Therefore, it is also important to establish that the load is not transferred to the hip following gait modification.

The purpose of this study was to determine how adjustments to velocity and step length affect knee joint loading in adults of healthy weight and those with obesity, a population at risk for knee OA. Specifically, peak tibiofemoral joint axial contact force, tibiofemoral joint axial impulse, and peak knee adduction angular impulse were compared among walking conditions of different step length and velocity combinations. We expected that slower velocity and shorter steps would reduce knee joint loading with greater effects when both factors were changed simultaneously. In addition, hip joint contact forces were evaluated for the possibility that knee joint compressive forces were redistributed to the hip.

METHODS

Level of evidence: Level 3 case-control study.

Participants

The university's Institutional Review Board provided approval prior to initiating the study. Young healthy adults between 18 and 45 years of age were recruited from the university's student body and the surrounding area. Those with a body mass index (BMI) between 18.5 and 24.9 (normal) or 30.0 and 40.0 (class I or II obesity) were eligible. Participants were excluded if they reported any contraindications from physical activity according to the Physical Activity Readiness Questionnaire (PAR-Q),[28] reported a previous lower extremity major injury, surgery, or arthritis, used a walking aid, or were unable to understand and follow instructions. Participants were recruited until each BMI group (healthy weight, obese) had 10 participants (five men, five women per group; Table 1). All participants provided written informed consent to participate prior to data collection. Sample size was estimated using G Power to detect a significant difference with $p < 0.05$ and 80% power for a two factor (step length, velocity) repeated measures on both factors analysis of variance.[29] A sample size of 10 participants was indicated to detect a significant interaction effect between velocity and step length with a moderate ($f = 0.25$) effect size.[29,30] The moderate effect size was chosen because it is equivalent to the reduction in knee adduction impulse reported with the use of laterally wedged in shoe orthoses, a current treatment option.[10] A significant interaction would indicate that the combined effect of step length and velocity modification was different than the sum of each effect individually.

Experiment Protocol

After providing his or her consent to participate, the participant's height and weight were measured without shoes using a stadiometer. BMI was calculated and the PAR-Q was completed to confirm that the participant met the criteria for the study. Participants then changed into laboratory-provided shorts, socks, and shoes in preparation for marker placement. Retroreflective tracking and anatomical markers were attached to the pelvis, lower limbs, and feet using double-sided skin safe tape and thermoplastic shells with neoprene underwraps according to standard procedures.[31] Briefly, anatomical markers were attached to the greater trochanters, medial and lateral epicondyles, medial and lateral malleoli, posteroinferior calcaneus, and first and fifth metatarsal heads. Tracking markers were affixed to shells located on the posterior pelvis, proximolateral thigh, posterodistal shank, and were attached directly to the posterosuperior, lateral, and medial aspects of the calcaneus. Marker position data were recorded at 100 Hz with an eight-camera motion capture system (Vicon, Oxford, UK). Two force platforms (AMTI Inc., Watertown, MA) sampling at 1000 Hz recorded ground reaction forces and were time synchronized with the motion capture system. Following collection of a standing trial for anatomical segment coordinate system determination, anatomical markers were removed.[32]

Participants initially completed five good trials of walking at their preferred speed in the gait laboratory for the baseline condition. A good trial was one in which participants contacted the force plate cleanly with each foot without any apparent adjustment to their gait. Walking velocity was

Table 1. Participant Demographic Characteristic for Each Group (Mean [Standard Deviation])

	Healthy Weight Group	Obese Group
Age (years)	26.5 (4.5)	26.6 (5.8)
Height (m)	1.68 (0.06)	1.69 (0.07)
Mass (kg)	62.93 (7.66)	96.86 (1.68)
BMI (kg/m^2)	22.2 (1.6)	33.7 (3.8)
Preferred velocity (m/s)	1.43 (0.14)	1.40 (0.17)
Preferred step length (m)	0.75 (0.05)	0.74 (0.07)

recorded using two photocells placed 6 m apart and attached to a timer (Brower Timing Systems, Draper, UT). Following completion of the baseline and subsequent conditions, participants reported their rate of perceived exertion (RPE) using a 14-point Borg scale.[33] The experimental conditions incorporated preferred (PV), 15% increased (IV), and 15% decreased (DV) walking velocities and preferred (PL), 15% increased (IL), and 15% decreased (DL) step lengths. The 15% change was calculated from the preferred condition. Including conditions altering both step length and velocity simultaneously, participants underwent eight experimental conditions, plus the preferred condition. Conditions are referred to by a combination of velocity and step length condition abbreviations, for example the baseline control condition is PV-PL. These conditions were presented to the participant in randomized order following the baseline condition. Velocity was adjusted by providing feedback to the participant based on output from timing gates. Adjustments to step length were achieved with the use of colored tape lines on the floor and a metronome indicating step frequency at the target velocity. Participants rested as much as desired between conditions.

Data Processing

Data for the right limb were processed in Visual 3D (C-Motion, Germantown, MD) using a six degree-of-freedom model and joint coordinate systems following the standardization recommendations of Cole et al.[32] The standing calibration trial data were used to establish segment anatomical coordinate systems. Three-dimensional joint angles and external joint moments were determined by resolving into the proximal segment. Marker trajectories and ground reaction force data were filtered using a low pass recursive Butterworth filter at 6 Hz.[34] Filter cut-off frequency was determined by residual analysis and selected to preserve 95% of the signal.[35] Stance phase was determined by a 20N threshold of the vertical ground reaction force to indicate foot strike and toe off.

In order to obtain knee and hip contact forces, stance phase angles were imported into a musculoskeletal model and implemented using a custom MatLab routine (Mathworks, Natick, MA) in the manner of Derrick et al.[36–38] Segment and muscle lengths of 44 muscles were scaled to individual anthropometrics.[39] Muscle moment arms, muscle orientations, and maximal dynamic muscle forces, adjusted for velocity and length, were derived using muscle parameters from Arnold et al.[39] Individual muscle forces across each stance phase were estimated using static optimization while minimizing the sum of muscle stresses squared. Muscle forces solutions were constrained as follows: (1) individual forces were not less than zero or greater than maximal muscle forces estimated from the musculoskeletal model; and (2) the moments caused by the muscle forces were equivalent to the joint moments calculated from inverse dynamics. Joint contact force at the knee and hip during each stance phase were then estimated using vector summation of the respective joint reaction forces and corresponding muscle forces crossing each joint. Variables of interest were extracted from the stance phase time series data and included peak tibiofemoral joint axial contact force and the impulse of this force. In addition, the peak resultant hip joint contact force was extracted for analysis. All joint forces and moments were normalized to fat free weight (FFW) and height. This normalization procedure reduces the variability in the data set due to variations in overall body size (lean body mass) and stature, but does not mask the effect of body fat on joint loads. FFW was estimated using a validated regression equation based on BMI, age, and sex.[40] Experimental walking data from Fregly et al.[41] were input to the model described above to estimate peak TFJ force and TFJ impulse. In general, curve profiles from our model prediction and in vivo tibia contact force data showed similar shape and timing (Figure 1). Furthermore, model estimates of trial data were within 4% and 5%, on average, of in vivo peak TFJ force, TFJ force impulse, respectively.

Additional descriptive variables of interest were extracted using custom Matlab programs (Mathworks, Natick, MA) and included peak external knee flexion and adduction moments and peak knee flexion and adduction angles during the first 60% of stance. Knee adduction angular impulse was calculated from the time integral of adduction moment during stance and extracted for statistical analysis. The mean of five trials for the discrete variables of interest were averaged per participant in each condition. Group means and standard deviations were then calculated and used for statistical analysis.

Two factor (step length and velocity) repeated measures on both factors analysis of variance was conducted separately for each group for peak tibiofemoral joint axial contact force, tibiofemoral joint axial impulse, and peak knee adduction angular impulse (SPSS 24, SPSS Inc., Chicago, IL). Each group was analyzed separately to determine the influence of changing velocity and step length on walking biomechanics compared to the PV-PL baseline control condition for the group. If Mauchly's test of sphericity was significant, the Greenhouse-Geisser correction was used. Post hoc least significant difference tests were used to compare each experimental condition with the preferred condition for significant main effects. In the case of a significant interaction effect, post hoc paired t-tests with Bonferroni correction ($p < 0.006$) were used for planned comparisons of each experimental condition with the preferred velocity-preferred

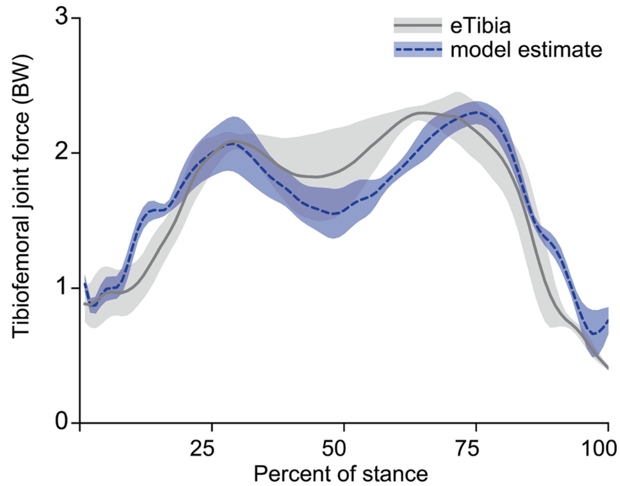

Figure 1. Average tibiofemoral joint contact forces for adults of healthy weight (gray lines) and those with obesity (black lines) while walking at A) −15% velocity, B) preferred velocity, and C) +15% velocity. Line types represent +15% step length (dash), preferred step length (solid), and −15% step length (dot). Shaded bands represent the range of peak tibiofemoral joint forces within each velocity condition. Impulse is the area under the curve.

step length baseline condition only. Thus, there were a maximum of eight planned post-hoc comparisons to determine differences from the PV-PL baseline control condition. The Bonferroni correction reflects this and divides the p value 0.05 by 8.

RESULTS

Participants were able to adjust walking velocity and step length in each of the experimental conditions (Supplementary Table S1). In the healthy weight group preferred walking velocity was 1.43 (0.14) ms^{-1} and preferred step length was 0.75 (0.05) m. In the obese group preferred walking velocity was 1.40 (0.17) ms^{-1} and preferred step length was 0.74 (0.07) m. There were no differences in RPE in the obese group among conditions and only a small (about 1 unit) increase in some conditions in the healthy weight group (Supplementary Table S2). Regarding knee joint contact force, the groups differed in whether responses were main effects for velocity and step length or an interaction effect. Peak tibiofemoral joint contact force had a significant interaction effect in the healthy weight group (Table 2; Figure 2; $p = 0.033$). IV-IL and PV-IL both increased peak force compared to baseline (by 26% and 2%). There were significant main effects for both velocity ($p < 0.001$) and step length ($p = 0.014$) in the obese group (Table 3; Figure 2) with no interaction ($p = 0.993$). DV reduced the peak force by 9% compared to baseline, and both IV and IL produced higher peak force in this group (by 10% and 8%).

Tibiofemoral joint impulse had significant main effects for velocity ($p < 0.001$) and step length ($p < 0.001$) in the healthy weight group with no interaction ($p = 0.106$). IV and DL both reduced the impulse (by 11% and 13%), while DV and IL both increased impulse compared to preferred (by 11% and 17%). The significant interaction effect for tibiofemoral joint impulse in the obese group ($p < 0.001$) indicated that both IV-DL and PV-DL reduced the impulse compared to baseline (by 19% and 12%). DV-PL, DV-IL, and PV-IL all increased the impulse (by 11%, 32%, and 13%).

Knee adduction angular impulse had significant main effects for velocity ($p < 0.001$) and step length ($p < 0.001$), but no interaction ($p = 0.279$) in the healthy weight group. Both IV and DL reduced the impulse compared to preferred (by 18% and 27%), while DV and IL increased the impulse (by 9% and 18%). There was a significant interaction for knee adduction impulse in the obese group ($p = 0.001$). All DL conditions plus IV-PL also reduced the impulse compared to baseline (by 21% and 14%).

Table 2. Primary Knee Joint Variables for Healthy Weight Participants in Each Condition (Mean [SD])

	Decreased Step Length	Preferred Step Length	Increased Step Length	All Step Lengths
Peak tibiofemoral joint contact force (N/FFW)				
Decreased velocity	2.93 (0.31)	3.14 (0.48)	3.50 (0.71)	3.19 (0.56)
Preferred velocity	3.34 (0.48)	3.32 (0.60)	3.76 (0.77)[c]	3.47 (0.64)
Increased velocity	3.29 (0.63)	3.74 (0.57)	4.18 (0.74)[c]	3.74 (0.73)
All velocities	3.19 (0.51)	3.40 (0.59)	3.81 (0.77)	
Tibiofemoral joint impulse (Ns/FFW)				
Decreased velocity	1.41 (0.14)	1.64 (0.16)	1.92 (0.13)	1.66 (0.25)[b]
Preferred velocity	1.31 (0.15)	1.46 (0.10)	1.72 (0.15)	1.50 (0.22)
Increased velocity	1.13 (0.12)	1.34 (0.13)	1.56 (0.15)	1.34 (0.22) [b]
All velocities	1.29 (0.18)[a]	1.48 (0.18)	1.73 (0.20)[a]	
Knee adduction impulse, Nms/(FFW × ht)				
Decreased velocity	0.010 (0.003)	0.013 (0.004)	0.015 (0.004)	0.012 (0.004)[b]
Preferred Velocity	0.008 (0.003)	0.011 (0.003)	0.013 (0.003)	0.011 (0.004)
Increased velocity	0.007 (0.002)	0.009 (0.002)	0.011 (0.003)	0.009 (0.003)[b]
All velocities	0.008 (0.003)[a]	0.011 (0.003)	0.013 (0.004)[a]	

[a]Significantly different than preferred step length ($p < 0.05$). [b]Significantly different than preferred velocity ($p < 0.05$). [c]Significant interaction between velocity and step length conditions ($p < 0.006$). Compressive forces are positive.

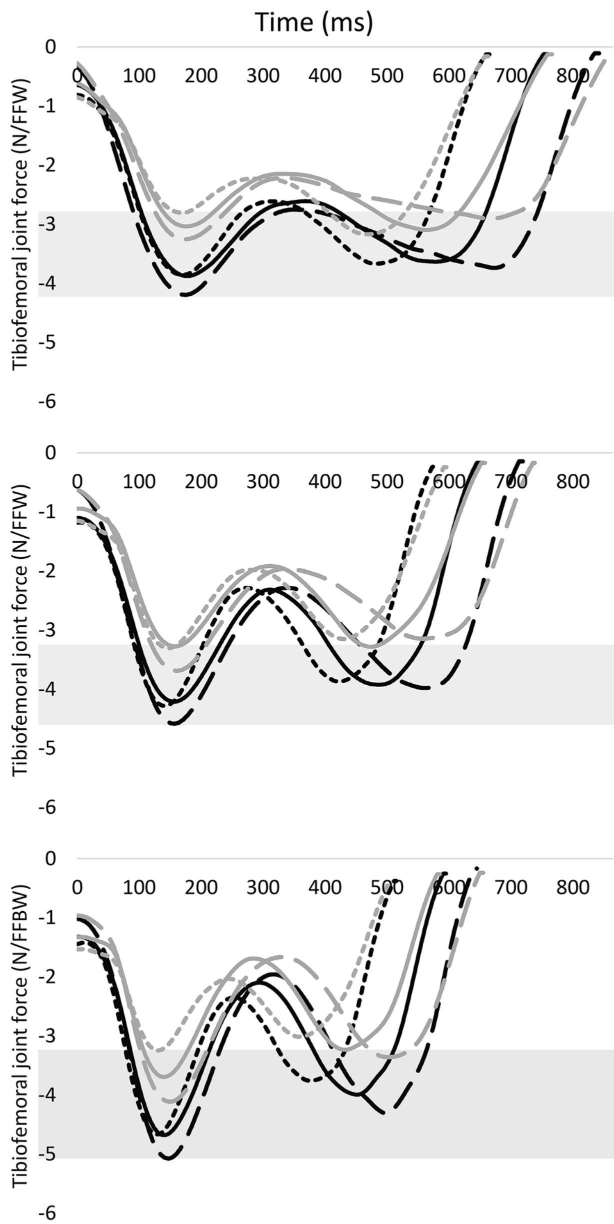

Figure 2. Tibiofemoral joint compressive force estimate using our musculoskeletal model and Fregly et al. (2012) input data, and the measured value from an instrumented prostheses (eTibia) recorded simultaneously during walking. The solid line represents the average eTibia measurements over five stance phases. The dashed line represents the average modeled tibiofemoral contact force during the same five trials. The shaded region represents one standard deviation above and below the mean.

Peak hip resultant joint contact force did not increase above baseline for any of the conditions which demonstrated a decrease in knee joint contact forces or kinetics in either the healthy weight or obese group (Supplementary Table S3). Similar patterns are observed for hip joint impulse (Supplementary Table S4). Furthermore, only small changes in peak knee joint moments and peak knee joint angles were noted in the experimental conditions compared to baseline (Supplementary Tables S5, S6).

DISCUSSION

The purpose of this study was to determine how adjustments to velocity and step length during walking affect knee joint loading in adults of healthy weight and those with obesity. Of particular interest were the adjustments which reduced knee joint loading compared to the baseline condition of preferred velocity and preferred step length. Changes were generally similar in both the adults of healthy weight and adults with obesity. Peak magnitudes of the variables of interest tended to decrease with decreases in velocity and step length, both alone and in combination. However, variables indicating the loading dose over the entire stance phase (impulses) tended to decrease with decreased step length, but to increase with decreased velocity, both alone and in combination. It is also important to note that the decreases in knee joint loads in the present study were not achieved by shifting the forces to the hip joint. Furthermore, the walking modifications were achieved without increasing perceived physical effort, as reflected in the minimal changes in the ratings of perceived exertion.

Manipulation of step length and gait velocity evoked changes in knee joint contact forces, particularly the tibiofemoral joint axial impulse. Changes in peak tibiofemoral joint axial contact force were small, and the only reduction was with decreased velocity in the adults with obesity. No decreases in peak force were found in the healthy weight group. Comparatively, the effects on tibiofemoral joint impulse were larger resulting in a 28% reduction in impulse in healthy weight and 12–19% reduction in the obese group with a 15% decreased step length. Interestingly, among adults with obesity, the potentially beneficial effect of decreased step length was dependent on maintaining preferred gait velocity. The combination of decreased step length and decreased velocity did not change tibiofemoral joint impulse from the preferred gait condition. This is potentially clinically relevant as decreased gait velocity is an organic effect of knee OA.[42] The finding that an acute change to a shorter step length during walking immediately reduces knee joint loading is in agreement with the predictions of a recent simulation study that sought to reduce knee joint forces.[17]

The percent decrease in tibiofemoral joint impulse is comparable to the percent decrease in peak joint contact force reported recently in a comparison of level walking with slow uphill walking in adults of healthy weight and adults with obesity.[7] In particular, first peak tibiofemoral joint contact force was decreased by 23%–35%. These reductions are larger than we found for peak force, but comparable to the reductions in joint contact impulse found with decreased step length. Adjusting velocity and step length during walking on level ground may provide a joint-healthy alternative to walking uphill on a treadmill, increasing the opportunities for adopting walking adjustments

Table 3. Primary Knee Joint Variables for Obese Participants in Each Condition (Mean [SD])

	Decreased Step Length	Preferred Step Length	Increased Step Length	All Step Lengths
Peak tibiofemoral joint contact force (N/FFW)				
Decreased velocity	4.00 (1.06)	3.95 (1.03)	4.31 (1.27)	4.08 (1.10)[b]
Preferred velocity	4.34 (1.12)	4.30 (1.02)	4.65 (1.49)	4.43 (1.20)
Increased velocity	4.73 (1.30)	4.74 (1.33)	5.12 (1.43)	4.86 (1.32)[b]
All velocities	4.36 (1.16)	4.33 (1.15)	4.69 (1.39)[a]	
Tibiofemoral joint impulse (Ns/FFW)				
Decreased velocity	1.81 (0.27)	2.09 (0.28)[c]	2.48 (0.39)[c]	2.13 (0.42)
Preferred Velocity	1.65 (0.26)[c]	1.88 (0.31)	2.12 (0.34)[c]	1.88 (0.36)
Increased Velocity	1.53 (0.25)[c]	1.75 (0.29)	1.96 (0.31)	1.74 (0.33)
All Velocities	1.66 (0.28)	1.91 (0.32)	2.19 (0.41)	
Peak knee adduction impulse, Nms/(FFW × ht)				
Decreased velocity	0.012 (0.005)[c]	0.016 (0.005)[c]	0.020 (0.006)[c]	0.016 (0.006)
Preferred velocity	0.011 (0.004)[c]	0.014 (0.005)	0.016 (0.004)[c]	0.013 (0.005)
Increased velocity	0.009 (0.003)[c]	0.012 (0.004)[c]	0.014 (0.004)	0.012 (0.004)
All velocities	0.011 (0.004)	0.014 (0.005)	0.017 (0.005)	

[a]Significantly different than preferred step length ($p < 0.05$). [b]Significantly different than preferred velocity ($p < 0.05$). [c]Significant interaction between velocity and step length conditions ($p < 0.006$). Compressive forces are positive.

during daily activities, in addition to during bouts of purposeful exercise. In addition, a recommendation for uphill walking may be less palatable to the target population than simply taking shorter steps while walking. Thus, we suggest adopting a shorter step length without reducing velocity during daily activity as a simple means to reduce the cumulative load on the knee joint articular surfaces during each step.

Step length and gait velocity adjustments also affected the knee adduction impulse and, to a lesser extent, the peak joint moments. Decreased step length conditions reduced knee adduction impulse. Knee adduction impulse is an indicator of cumulative load per step and could be used as the basis for an accumulated daily load if individual step counts were monitored. However, the potentially beneficial effect of a 21-27% overall reduction in knee adduction impulse per step may be partially diminished to by the need for a greater number of steps required to walk a given distance following a 15% decrease in step length.[14] However, it should be noted that interventions to reduce these kinetic variables have not yet been found to reduce the incidence of knee OA or slow the rate of progression of established OA in prospective, longitudinal studies. Nevertheless, within the context of current knowledge, the joint contact force results in this study support step length reductions as a beneficial adjustment to daily walking activity even though there may be a slight metabolic cost to suggested modifications. While not measured in the present study, decreasing step length by 15% is reported to increase metabolic cost 4.6% during treadmill walking.[18]

The magnitude of changes in knee joint loading achieved by decreasing step length are comparable to changes in knee joint biomechanics implemented by different gait modifications or by orthopedic devices. Ipsilateral trunk lean was effective in decreasing knee adduction impulse (by 35%),[9] but the 11% increase in energy expenditure and large cosmetic change required may not be acceptable to the target population. The decrease in knee adduction impulse reported for the toe out gait style[9] (14%) was less than the decreases observed for our comparable healthy weight individuals (27%) during the walking modifications in the present study. Although, toeing out did not greatly increase energy expenditure (2%) either and was thus broadly similar in effect to our proposed walking modifications. Modest reductions in knee adduction impulse following a 2-week period of using either a valgus knee brace (9%) or laterally wedged insoles (16%) in overweight and obese adults with knee osteoarthritis have been reported,[10] slightly below our mean reduction of 21% in obese young adults who decrease step length. However, patients with a prescription for knee braces have a low level of adoption.[43] Given the greater patient acceptance of insoles and larger decreases in impulse reported relative too bracing,[10] this mechanical intervention may be broadly similar in effect to our proposed walking modifications.

It should be noted that all joint contact forces were estimated from a musculoskeletal model and not measured directly. The main findings of this study were based on within-subject comparisons, which should minimize influences of model assumptions and error on condition differences. For ease of comparison to prior published work, peak TFJ force and TFJ impulse for health weight controls in this study averaged 2.78–3.07 BW and 0.30–0.33 BWs, respectively, for walking at speeds ranging from 1.23–1.64 m/s. Peak medial and lateral TFJ force averaged 2.05–2.33 BW and 0.72–0.87 BW, respectively. Comparatively at speeds ranging from 0.70–1.39 m/s, peak tibiofemoral contact force measured in vivo during over ground walking is reported to range from 1.8–3.0 BW, with medial force ranging from 1.2–2.0 BW and lateral force ranging from 0.5–1.0 BW.[41] Average peak resultant hip contact force in this study ranged from 3.57–3.73 BW across walking speeds ranging from 1.23–1.64 m/s for healthy weight controls. Whereas in vivo studies report peak hip contact forces ranging from 2.5–4.7 BW for walking at speed ranges of 1.4–1.8 m/s.[44–46] Based on these comparisons, the model for this study appeared to produce hip and knee contact force outcomes that are generally in the range of in vivo data. Combined with the direct comparison to in vivo knee forces described in the methods, the model used in this study demonstrates good fidelity when compared to in vivo measurements.

This study is limited by the acute nature of the walking adjustments. Participants performed all eight experimental gait adjustments in a single session. However, significant decreases in biomechanical risk factors for knee OA were found in both the healthy weight and obese groups. The long-term effects of these walking adjustments must be determined in future studies. Furthermore, the implementation of the changes in step length in the laboratory cannot be adopted directly for walking in the outdoor environment. Future work should focus on developing the means to decrease step length while maintaining or decreasing velocity during every day walking.

In summary, the acute effects of changes in walking velocity and step length on knee joint loads were determined in both healthy weight adults and adults with obesity. Reductions in peak TFJ force and TFJ impulse were achieved by decreasing step length during walking. Decreasing gait velocity, on the other hand, reduced peak TFJ force but increased TFJ impulse. Ratings of perceived exertion indicated that these walking modifications are likely to be acceptable to both healthy weight adults and adults with obesity. Our findings provide initial evidence that step length and gait velocity manipulations may have a role to play in knee OA prevention efforts. In particular, decreasing step length can reduce knee joint loads during daily walking and walking for exercise in people at risk for disability due to knee OA.

AUTHORS' CONTRIBUTIONS

CEM—research design, analysis and interpretation of data, drafting and critically revising the paper; SAM—analysis and interpretation of data, critically revising the paper; JLH—analysis and interpretation of data, drafting the paper; JDW—analysis and interpretation of data, critically revising the paper. All authors have read and approved the final submitted manuscript.

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SUPPORTING INFORMATION

Additional supporting information may be found online in the Supporting Information section at the end of the article.