Gait & Posture 70 (2019) 24–29

Contents lists available at ScienceDirect

Gait & Posture

journal homepage: www.elsevier.com/locate/gaitpost

# Physical activity and age-related biomechanical risk factors for knee osteoarthritis

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ARTICLE INFO

*Keywords:*
Muscle strength
Muscle power
Older gait
Co-activation
Knee mechanics

ABSTRACT

*Background:* Knee osteoarthritis (OA) is a highly prevalent disease leading to mobility disability in the aged that could, in part, be initiated by age-related alterations in knee mechanics. However, if and how knee mechanics change with age remains unclear.
*Research question:* What are the impacts of age and physical activity (PA) on biomechanical characteristics that can affect the loading environment in the knee during gait?
*Methods:* Three groups (n = 20 each, 10 male and 10 female) of healthy adults were recruited: young (Y, 21–35 years), mid-life highly active (MHi, 55–70 years, runners), and mid-life less active (MLo, 55–70 years, low PA). Outcome measures included knee kinematics and kinetics and co-activation during gait, and knee extensor muscle torque and power collected at baseline and after a 30-minute treadmill trial to determine the impact of prolonged walking on knee function.
*Results:* At baseline, high-velocity concentric knee extensor power was lower for MLo and MHi compared with Y, and MLo displayed greater early (6.0 ± 5.8 mm) and peak during stance (11.3 ± 7.8 mm) femoral anterior displacement relative to the tibia compared with Y (0.2 ± 5.6 and 4.4 ± 6.8 mm). Also at baseline, MLo showed equal quadriceps:hamstrings activation, while Y showed greater relative hamstrings activation during midstance. The walking bout induced substantial knee extensor fatigue (decrease in maximal torque and power) in Y and MLo, while MHi were fatigue-resistant.
*Significance:* These results indicate that maintenance of PA in mid-life may impart small but measurable effects on knee function and biomechanics that may translate to a more stable loading environment in the knee through mid-life and thus could reduce knee OA risk long-term.

## 1. Introduction

Knee osteoarthritis (OA) is a mobility-limiting, age-related disease for which there is a 44% lifetime risk in American adults [1]. Maintenance of physical activity (PA) throughout the lifespan may reduce OA risk [2], possibly by mitigating changes in several knee OA biomechanical risk factors that are also associated with aging (e.g., decreased quadriceps strength, increased muscle co-activation, altered knee mechanics during gait) [3–5]. As rates of knee OA initiation increase rapidly from age 50–70 years [6], quantifying age-related changes in biomechanical OA risk factors in mid-life, and the potential for PA to mitigate them, is critical.

There is initial evidence that healthy adults at mid-life or older walk with less knee flexion at heel strike and a smaller range of motion at the knee [7], greater knee flexion in midstance [8], and greater femoral anterior displacement relative to the tibia [9] compared with young adults. These differences are also characteristic of symptomatic knee OA [9,10]. The co-occurrence of the altered knee mechanics with age and knee OA suggests these changes may precede the onset of symptoms thereby supporting a mechanical pathway for idiopathic knee OA initiation [11]. However, as highlighted in a recent meta-analysis, few studies have quantified the age-related changes in knee mechanics and if, how, and in whom these changes occur remains unclear [7].

Abnormal knee mechanics could partially result from decreased muscle strength. Around mid-life, decreases in knee extensor muscle torque and power [12] and increased knee extensor fatigue following dynamic contractions begin to occur [13]. Decreased knee extensor function is associated with knee OA initiation [14] and altered knee flexion angles during gait in individuals at risk of knee OA post-ACL rupture [15] or with current knee OA [16]. Because low knee extensor

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https://doi.org/10.1016/j.gaitpost.2019.02.008
Received 15 November 2018; Received in revised form 15 January 2019; Accepted 12 February 2019
0966-6362/ 

strength is itself a risk factor for knee OA, further loss of strength with fatigue could amplify knee OA risk. Changes in knee mechanics may also result from altered muscle activation. Initial evidence suggests mid-life adults with and without knee OA have greater muscle activation across the knee compared to young adults [17], which could alter knee joint loading [18].

If changes in strength or muscle activation drive age-related changes in knee mechanics, we would expect greater deviations in knee mechanics in adults with poorer knee extensor function. Higher PA is associated with greater knee extensor muscle torque and power in mid-life and older adults [19], and has been shown to not increase the risk [20] and possibly protect against [2] and slow the progression of knee OA [21]. However, there is a lack of information on the role of PA in biomechanical risk factors for knee OA, specifically, knee extensor muscle strength and fatigability and muscle co-activation and knee mechanics during gait.

This study's **primary aim** was to determine whether knee extensor muscle function (here, maximal torque and power), co-activation across the knee during gait, and knee mechanics differ between young (Y) and mid-life adults, and between mid-life adults with high (MHi) or low (MLo) PA levels. We hypothesized that MLo would be weaker, have altered muscle co-activation, and display different knee mechanics compared to Y and MHi. Further, even if individuals are similar at baseline, the potential for daily bouts of activity to induce muscle fatigue may predispose certain populations to fatigue-related changes in knee mechanics or muscle activation and, potentially, increased knee joint loads. Therefore, our **secondary aim** was to test the impact of an acute exercise bout on knee extensor muscle function and knee mechanics during gait. We hypothesized MLo would display greater knee extensor fatigue in response to a bout of walking, and have correspondingly greater changes in knee mechanics and co-activation compared to both Y and MHi.

## 2. Methods

### *2.1. Participant selection*

Three groups were recruited: highly active mid-life adults (MHi; 55–70 years, running ≥15 miles/wk), less active mid-life adults (MLo; 55–70 years, ≤3 30-minute moderate exercise bouts/wk), and young adults (Y; 21–35 years; recreationally active). Groups included equal male and female numbers. MHi were runners to ensure a vigorously active group that could be quantified using accelerometry. Y were recreationally active (but not regular runners) as this activity profile matched that of highly active mid-life adults in preliminary data collection. All participants completed Knee Osteoarthritis Outcome Score questionnaires to verify absence of knee symptoms. Scores were similar between groups (Supplementary Table S-1). All participants had BMI < 30 $kg{\cdot}m^{-2}$, were free of significant musculoskeletal injury history, cardiovascular or neurological pathology, and chronic pain. Power calculations indicated 12–19 participants per group were needed to detect meaningful differences (Supplementary Table S-2). Participants completed IRB-approved informed consent prior to data collection.

### *2.2. Study protocol*

Participants completed two study visits ≥7 days apart. Visit one included a timed 400 m walk at self-selected pace to determine treadmill speed for visit two (see *30-minute treadmill walk*). Participants also practiced strength testing and were given a PA monitor. Visit two included overground walking gait analysis; knee extensor muscle testing; and a 30-minute treadmill walk (30MTW) with electromyography (EMG) collection. During this visit, participants wore standard neutral shoes (RC550, New Balance, Boston, MA).

#### *2.2.1. Physical activity monitoring*

All participants wore triaxial accelerometers (GT3X, Actigraph, Pensacola, FL) at the hip for 7 days. PA data included ≥4 days of ≥10 h wear, including ≥1 weekend day. Accelerometer data were used to calculate average weekly activity counts and moderate-to-vigorous PA (MVPA) minutes [22].

#### *2.2.2. Gait analysis*

Overground gait was captured before and after the 30MTW. Kinematics and kinetics of participants' right leg were captured using an 11-camera motion analysis system (Oqus, Qualisys, Göteborg, Sweden) with 2 force plates (AMTI, Watertown, MA). Marker and force data were collected at 200 and 2000 Hz, and low-pass filtered at 8 and 15 Hz, respectively. Five acceptable trials were captured at each of 2 speeds: preferred and fixed (1.4 $m{\cdot}s^{-1}$). Acceptable trials involved the participant cleanly hitting a force plate with their right foot at a speed within 5% of the other trials for that condition (monitored via photogates).

Thigh and shank segments were modeled using the Point Cluster Technique (PCT). PCT is a previously-validated [23,24] marker configuration and algorithm optimized for calculation of the 3 rotations and 3 translations at the knee joint using clusters of markers on the thigh (10 markers) and shank (7 markers). Pelvis, thigh, shank and foot coordinate systems were established during a static trial from anatomic markers (anterior and posterior iliac spine, iliac crest, greater trochanter, medial and lateral femoral epicondyles, medial and lateral tibial plateau, medial and lateral malleoli, calcaneus and 5th metatarsal). The foot and pelvis were tracked by their anatomic markers. Externally-referenced joint moments were calculated using inverse dynamics.

Knee kinematic outcomes included flexion angle at heel strike, midstance peak and range of motion during stance; peak adduction angle during loading response; and femoral anterior displacement relative to the tibia at heel strike, at the first peak of the vertical ground reaction force, average over stance, and peak during stance. Knee kinetics included the first peak extension and adduction moments, and the peak flexion moment. For descriptive purposes, sagittal hip and ankle ranges of motion during stance and peak flexion and/or extension moments were also reported.

#### *2.2.3. Knee extensor muscle function testing*

Maximal isometric torque ($Nm{\cdot}kg^{-1}$) as well as peak concentric and eccentric isokinetic knee extensor power ($W{\cdot}kg^{-1}$) at 90 and 270$^{\circ}{\cdot}s^{-1}$ were collected before and after the 30MTW using an isokinetic dynamometer (HUMAC NORM, CSMi, Stoughton, MA). Concentric and eccentric power were collected in a single motion. At baseline, two sets of three repetitions were performed for each test (isometric, con/eccentric at 90$^{\circ}{\cdot}s^{-1}$, con/eccentric at 270$^{\circ}{\cdot}s^{-1}$) with 30 s rest between sets and 2 min rest between tests. Isometric repetitions included 5 s contractions followed by 5 s rest. After the 30MTW one set of each test was collected with 15 s rest between tests. For isometric torque, the knee was flexed 60° relative to full extension and isokinetic power was collected across 70° of knee motion.

#### *2.2.4. 30-minute treadmill walk (30MTW)*

After baseline gait and strength testing, participants performed the 30MTW. Treadmill speed was set to the pace of the 400 m walk from visit one. If this pace was not comfortable, treadmill speed was adjusted in increments of 0.1 mph until the speed felt "normal." Treadmill incline was increased to 3% at minutes 7, 17, and 27 for one minute and then returned to level. This protocol was designed to mimic 30 min of exercise an individual might complete during a typical day, and has been shown to cause knee extensor fatigue in older women [25].

#### *2.2.5. Knee muscle co-activation*

Co-activation was calculated using surface EMG collected at 2000 Hz. Electrodes (Trigno, Delsys, Natick, MA) were placed on the rectus femoris, vastus lateralis, vastus medialis, biceps femoris, and semitendinosus. Ten consecutive strides were extracted from the second and final minutes of the 30MTW. Gait events were identified using an accelerometer on the lower leg. Each signal had the mean offset removed and was band-pass filtered at 20–500 Hz, rectified, and lowpass filtered at 20 Hz. Each muscles' resulting signal was then normalized to its average stance activation over the 10 strides during the second minute of the 30MTW.

Directed co-contraction ratios (DCCRs) were calculated between the quadriceps (average of rectus femoris and vasti) and hamstrings (average of biceps femoris and semitendinosus) [10]. DCCRs were calculated at each gait cycle point *t* for each stride *s*:

If quadriceps activation > hamstrings activation:

$$DCCR_{t,s} = 1 - \frac{(average\ \ of\ \ hamstrings\ \ linear\ \ envelopes)_{t,s}}{(average\ \ of\ \ quadriceps\ \ linear\ \ envelopes)_{t,s}}$$

else,

$$DCCR_{t,s} = \frac{(average\ \ of\ \ quadriceps\ linear\ \ envelopes)_{t,s}}{(average\ \ of\ \ hamstrings\ linear\ envelopes)_{t,s}} - 1$$

DCCRs range between 1 and -1 where 1 indicates exclusive quadriceps and -1 exclusive hamstrings activation. Values near 0 indicate relatively equal activation of the two muscle groups. DCCRs were averaged across the 10 strides from the second and final minute of the 30MTW, and then over specific phases of the gait cycle: terminal swing (final 15% of swing); and early, mid, and late thirds of stance.

#### 2.2.6. Statistics

Data were normally distributed, thus primary outcome variables were compared between groups using one-way MANOVAs with significance set at $p \leq 0.05$. For the primary aim, outcome variables were compared between groups at baseline (baseline overground gait and muscle strength, co-activation from 30MTW minute 2). For the secondary aim, changes in outcome variables pre- to post-30MTW were compared between groups (post-pre overground gait and muscle strength, minute 30-minute 2 co-activation data). Where significant main effects were found, Tukey's post-hoc tests were performed. Baseline and post-30MTW knee extensor torque and power were compared using paired t-tests to test for knee extensor muscle fatigue.

## 3. Results

Group characteristics are shown in Table 1. Due to technical issues, muscle co-activation data from the second minute of the 30MTW include 58 participants (n = 18 MLo) and for the final minute include 54 participants (n = 18, 19, and 17 Y, MHi, and MLo, respectively). MLo had fewer MVPA minutes compared to Y and MHi, and all groups differed in PA counts (Table 1).

### 3.1. Baseline comparison

At baseline, groups differed in knee extensor power at 90°·s$^{-1}$ (p = 0.01) and 270°·s$^{-1}$ (p = 0.002). Knee extensor power was lower in MLo compared to Y during concentric contractions at 90°·s$^{-1}$ (post-hoc p = 0.01) and in both MHi and MLo compared to Y at 270°·s$^{-1}$ (p = 0.006 for Y vs. MHi and MLo; Fig. 1A). During the second minute of the 30MTW groups differed in co-activation at midstance (p = 0.03), where MLo had greater quadriceps:hamstrings co-activation compared to Y (post-hoc p = 0.04; MLo DCCR = -0.01, Y DCCR = -0.22, Fig. 2A).

Baseline knee mechanics were similar between groups at both walking speeds. The fixed speed results are presented here (Table 2) with preferred speed results in supplements (Tables S-3, S-4). MLo had greater femoral anterior displacement than Y at the time of the first peak of the vertical ground reaction force, and greater peak femoral anterior displacement. Knee joint angles did not differ between groups (Fig. 3, Table 2). Y had greater knee extension moments in early stance compared to both MHi and MLo, and there was a trend towards MHi having larger knee flexion and first peak adduction moments than Y and MLo (Table 2).

### 3.2. Response to 30MTW

The 30MTW elicited knee extensor fatigue (decrease in torque or power) across most contraction velocities in MLo and Y but not in MHi (Fig. 1B). MLo fatigued more than MHi in concentric contractions at 270°·s$^{-1}$ (post-hoc p = 0.008, Fig. 1B). During terminal swing, MHi had a decrease vs. MLo's increase in quadriceps:hamstrings co-activation, (post-hoc p = 0.05, Fig. 2B).

Changes in knee kinematics after the 30MTW were small (< 1.5° or < 1 mm) and not different between groups (Table 3). Knee flexion moments changed differently between groups (Table 3), with MHi displaying a small decrease and Y a small increase in response to the 30MTW (post-hoc p = 0.03).

## 4. Discussion

The overall study aim was to determine the impact of age and PA on measures that may affect the loading environment in the knee, and thus knee OA risk, during gait. We hypothesized that MLo compared to both MHi and Y would have greater knee flexion angles and femoral anterior displacement during stance, altered knee moments, lower knee extensor muscle torque and power, and greater muscle co-activation. In partial support of this hypothesis, MLo were weaker than Y in concentric contractions, and had greater femoral anterior displacement and midstance co-activation patterns compared to MHi and Y. Further, we hypothesized that these group differences would be amplified by a 30MTW. This hypothesis was largely not supported, but MLo showed greater decreases in high-velocity knee extensor power than MHi. Our results suggest that high levels of PA in mid-life may mitigate mechanical risk factors for knee OA, particularly concentric knee extensor muscle power and femoral anterior displacement.

Similar to previous research [13], baseline knee extensor power and torque differed by age during concentric contractions. As expected, only MLo were weaker than Y at the moderate contraction velocity of 90°·s$^{-1}$, while both mid-life groups were weaker than Y at 270°·s$^{-1}$. The lack of a group difference in eccentric power aligns with studies comparing young and older (> 65 years) adults that have found smaller [26] and later [27] declines in eccentric relative to concentric knee extensor power with age. MLo's activity level of nearly 150 min per week of MVPA appeared to not be sufficient to preserve low-velocity

**Table 1**
Group characteristics reported as Mean (SD). MVPA: moderate to vigorous physical activity. Y: young group. MHi: highly active mid-life group. MLo: less active mid-life group. * different from Y; + different from MHi.

| Group | n (#male) | Age (years) | Height (m) | Mass (kg) | Preferred walking speed (m·s$^{-1}$) | Treadmill walking speed (m·s$^{-1}$) | Weekly MVPA minutes | Weekly counts (x10$^{-3}$) |
|---|---|---|---|---|---|---|---|---|
| Y | 20 (10) | 27.8 (3.5) | 1.72 (0.09) | 69.8 (11.8) | 1.40 (0.15) | 1.39 (0.14) | 393.5 (162.0) | 2509 (783) |
| MHi | 20 (10) | 61.9 (4.0) | 1.68 (0.11) | 64.4 (12.9) | 1.35 (0.12) | 1.35 (0.12) | 473.5 (216.5) | 3340 (1152) * |
| MLo | 20 (10) | 62.9 (3.9) | 1.71 (0.11) | 69.9 (11.7) | 1.35 (0.12) | 1.35 (0.12) | 147.7 (110.1) * + | 1504 (633) * + |

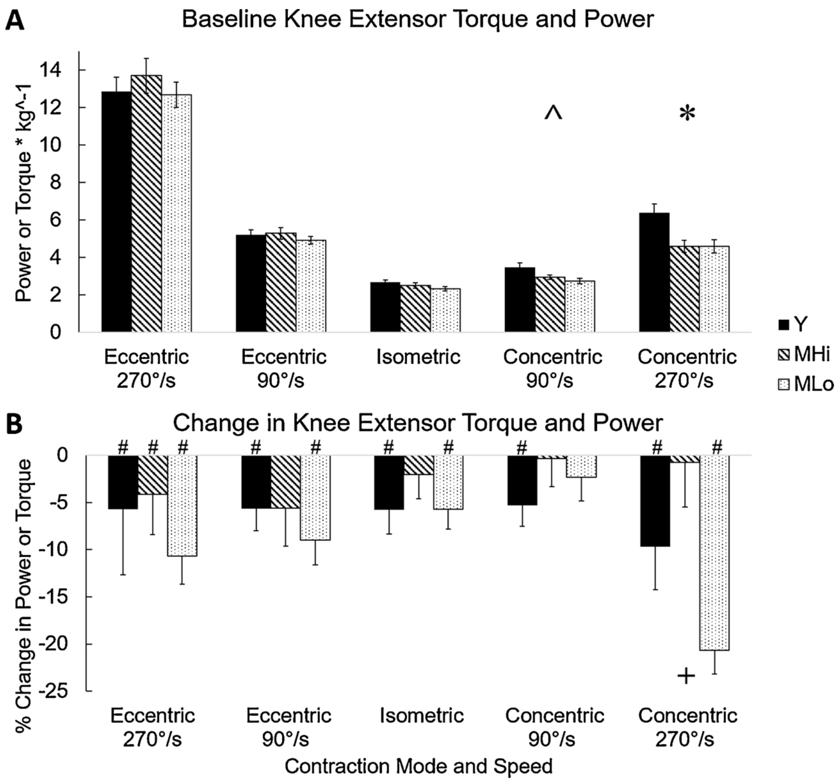


**Fig. 1.** Knee extensor torque and power at baseline (**A**) and change after the 30 min treadmill walk (**B**). Mean ± SE. Y: young group. MHi: highly active older group. MLo: less active older group. A: ^ indicates MLo different from Y; * indicates MHi and MLo different from Y. B: # indicates significant decrease from baseline. + indicates greater change in MHi than MLo.

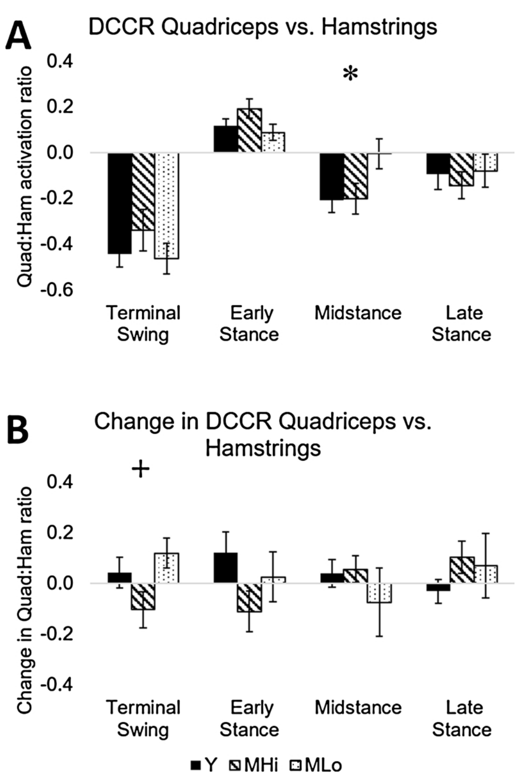


**Fig. 2.** DCCR for quadriceps vs. hamstrings at baseline (**A**) and change after the 30 min treadmill walk (**B**). Mean ± SE. Positive values indicate greater quadriceps activation relative to hamstring activation. Values near 0 indicate relatively equal activation between the two muscle groups. A: * indicates MLo different from Y. B: + indicates MHi different from MLo.

concentric muscle strength. As decreased knee extensor strength at moderate contraction velocities (60-120°·s$^{-1}$) has been associated with knee OA incidence [5], our results suggest that MLo may be at greater risk of knee OA than MHi in terms of knee extensor power.

At baseline, only MLo had greater quadriceps:hamstrings co-activation at midstance compared to Y. This could indicate that the quadriceps remain activated longer after early stance in MLo relative to Y and MHi. This finding is similar to the increased co-activation previously reported in older compared to young adults [28] and may suggest an earlier onset of age-related gait changes in MLo. While external joint moments and muscle activation do not directly assess loads in the knee, the combination of trends toward age-related increases in peak knee moments and greater co-activation would be consistent with higher contact forces in the knee in MLo [10,18]. MLo also had greater femoral anterior displacement compared to Y. Femoral anterior displacement appears to increase from young asymptomatic adults to mid-life and older asymptomatic adults and adults with symptomatic knee OA [9]. A combination of greater femoral anterior displacement with greater co-activation and knee moments could transfer higher loads to unconditioned cartilage [11], thereby increasing risk for knee OA initiation. These baseline results suggest that less active mid-life adults may be progressing along a trajectory towards knee OA initiation, and that, in agreement with recent literature [21], high levels of PA may protect against OA via an impact on biomechanical risk factors for knee OA.

The 30MTW induced knee extensor fatigue in MLo and, somewhat surprisingly, in Y, with variation in the magnitude of this effect across contraction velocities. The lack of concurrent changes in co-activation and gait mechanics suggests a relative insensitivity of gait mechanics to moderate fatigue in healthy young and middle aged adults. Based on limited literature, older mobility-intact adults may use only ≈25% of their knee extensor power during gait [29]. With a conservative assumption that MLo had the same relative effort as older adults in the literature, their ≈20% decrease in power would still leave a ≈55% reserve of knee extensor strength for maintaining gait.

The current study has some limitations. While MLo were less active than MHi and Y, their MVPA of ≈148 min·wk$^{-1}$ approached minimum exercise guidelines and exceeded reported averages for adults age 50+ [30]. A less active group may have resulted in larger differences both at

**Table 2**
Baseline knee kinematics and kinetics. KF: knee flexion; KA: knee adduction; FAD: femoral anterior displacement; KE: knee extension; HF: hip flexion; HE: hip extension; ADF: ankle dorsiflexion; Y: young group; MHi: mid-life highly active group; MLo: mid-life less active group. Where significant main effects were found, data are bolded and post-hoc p-values are reported. * indicates Y different from MLo, ˆ indicates Y different from MHi.

| | Y | | MHi | | MLo | | p-value | post-hoc |
|---|---|---|---|---|---|---|---|---|
| | Mean | SD | Mean | SD | Mean | SD | | |
| Stride length (m) | 1.49 | 0.06 | 1.45 | 0.10 | 1.47 | 0.08 | 0.22 | na |
| KF Heel strike (°) | 5.3 | 4.6 | 4.9 | 4.5 | 6.1 | 5.6 | 0.77 | na |
| KF Midstance (°) | 20.1 | 5.2 | 22.8 | 6.0 | 22.8 | 5.2 | 0.20 | na |
| KF Stance ROM (°) | 38.9 | 3.5 | 38.6 | 4.1 | 40.4 | 3.8 | 0.28 | na |
| KA Midstance (°) | 2.0 | 2.7 | 1.3 | 3.4 | 1.2 | 3.4 | 0.74 | na |
| FAD Heel strike (mm) | −9.0 | 7.1 | −6.0 | 7.5 | −5.2 | 7.0 | 0.23 | na |
| FAD at first VGRF peak (mm) | **0.2** | **5.6** | **3.9** | **4.9** | **6.0** | **5.8** | < **0.01** | * < 0.01 |
| FAD Stance average (mm) | **2.7** | **4.0** | **5.8** | **4.5** | **6.1** | **5.3** | **0.04** | na |
| FAD Max stance (mm) | **4.4** | **6.8** | **8.4** | **7.2** | **11.3** | **7.8** | **0.02** | *0.01 |
| KE Moment (%BW·Ht) | **−1.9** | **0.4** | **−1.5** | **0.6** | **−1.5** | **0.5** | **0.02** | *0.03, ˆ0.04 |
| KF Moment (%BW·Ht) | 2.6 | 1.3 | 3.5 | 1.4 | 2.8 | 1.0 | 0.08 | na |
| KA Moment (%BW·Ht) | −2.9 | 0.7 | −3.4 | 0.6 | −3.1 | 0.8 | 0.07 | na |
| Hip ROM (°) | 42.9 | 5.1 | 44.5 | 4.2 | 45.4 | 5.5 | | |
| Ankle ROM (°) | 26.1 | 4.0 | 26.1 | 3.9 | 28.3 | 4.7 | | |
| HF Moment (%BW·Ht) | −3.8 | 0.7 | −4.3 | 0.9 | −4.0 | 1.3 | | |
| HE Moment (%BW·Ht) | 4.5 | 1.0 | 4.5 | 1.1 | 4.0 | 1.3 | | |
| ADF Moment (%BW·Ht) | −9.3 | 0.7 | −9.1 | 0.7 | −8.9 | 1.3 | | |

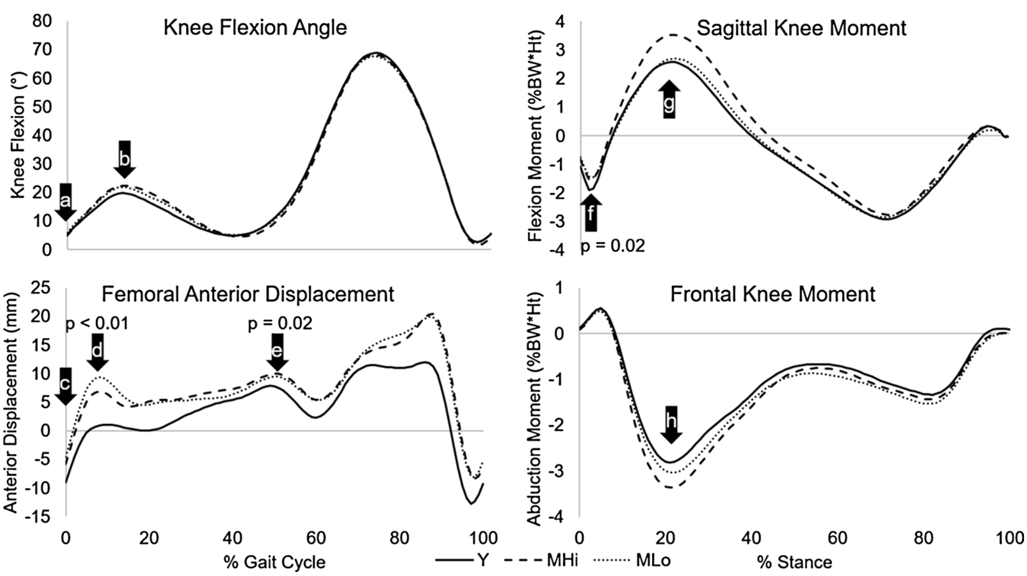


**Fig. 3.** Group mean knee kinematic and kinetic data through the gait cycle. Y: young group; MHi: mid-life highly active group; MLo: mid-life less active group. Lettered arrows indicate discrete variables of interest, reported in Table 2: a: knee flexion at heel strike; b: knee flexion at midstance; c: femoral anterior displacement (FAD) at heel strike; d: FAD at first vertical ground reaction force peak; e: FAD maximum during stance; f: first peak knee extension moment; g: peak knee flexion moment; h: first peak knee adduction moment. P-values noted where p ≤ 0.05 for main comparisons of group.

baseline and in response to the 30MTW. However, these mid-life groups had similar weekly minutes in light to moderate PA (2252 ± 566 for MHi and 1893 ± 555 for MLo), suggesting that differences observed are a result of the vigorous activity performed by MHi. Our ability to test that MLo would display greater fatigue *and* larger changes in gait mechanics than Y and MHi may have been limited by the moderate fatigue we observed in MLo. Finally, as with many studies of asymptomatic adults, we did not have radiographs and cannot exclude the possibility of asymptomatic knee OA.

Readers should be aware of the limitations of methodologies for calculating joint kinematics and the implications of a methodology relative to the hypotheses being tested. We selected PCT to measure tibiofemoral motion as our study questions necessitated that participants walk naturally at their preferred gait speed overground. In a PCT validation study using a tibial Ilizarov frame [23], average bone location error was 0.08 mm. If similar error were assumed for femur location, maximum error expected in tibia vs. femur location would be 0.16 mm, well below the group differences reported in our study.

We have shown that variables that may reflect the loading environment in the knee joint (i.e., anterior displacement of the femur relative to the tibia, midstance muscle co-activation, and knee extension moment in early stance) differ by age and habitual PA. Our results suggest high PA mitigates some age-related biomechanical risk factors for knee OA in healthy mid-life adults. The well-controlled cohorts in this study allowed for discrimination of factors that could alter the loading environment for knee joint cartilage based on age or decreased PA alone, independent of the many comorbidities that additionally alter cartilage health. Our results support a role of PA (independent of its tendency to reduce body weight and associated health problems) in wellness interventions or rehabilitation programs designed to reduce risk factors of knee OA.

## Conflict of interest statement

The authors declare no conflicts of interest.

## Acknowledgements

The authors thank Marquis Hawkins, PhD for statistical guidance. This project was supported by an American College of Sports Medicine

**Table 3**
Changes in knee kinematics and kinetics in response to the 30MTW. KF: knee flexion; KA: knee adduction; FAD: femoral anterior displacement; KE: knee extension; HF: hip flexion; HE: hip extension; ADF: ankle dorsiflexion; Y: young group; MHi: mid-life highly active group; MLo: mid-life less active group. Where significant main effects were found, data are bolded and post-hoc p-values are reported. * indicates Y different from MLo, ^ indicates Y different from MHi.

| | Y | | MHi | | MLo | | p-value | post-hoc |
|---|---|---|---|---|---|---|---|---|
| | Mean | SD | Mean | SD | Mean | SD | | |
| Stride length (m) | 0.00 | 0.04 | 0.00 | 0.02 | 0.01 | 0.04 | 0.34 | na |
| KF Heel strike (°) | 0.2 | 2.5 | 1.1 | 1.9 | 1.1 | 1.7 | 0.31 | na |
| KF Midstance (°) | 0.0 | 1.9 | 0.0 | 2.0 | 0.5 | 1.7 | 0.61 | na |
| KF Stance ROM (°) | −0.4 | 1.5 | −1.4 | 2.7 | −1.1 | 1.8 | 0.29 | na |
| KA Midstance (°) | −0.3 | 1.0 | −0.4 | 1.3 | −0.5 | 1.3 | 0.90 | na |
| FAD Heel strike (mm) | −0.8 | 3.8 | 0.4 | 4.0 | −0.3 | 5.0 | 0.69 | na |
| FAD at first VGRF peak (mm) | −0.6 | 3.7 | 0.0 | 4.2 | −0.3 | 4.7 | 0.88 | na |
| FAD Stance average (mm) | −0.7 | 3.7 | −0.4 | 3.8 | 0.1 | 3.9 | 0.83 | na |
| FAD Max stance (mm) | −0.1 | 3.9 | 0.2 | 4.4 | 0.3 | 3.8 | 0.64 | na |
| KE Moment (%BW·Ht) | −0.2 | 0.3 | 0.0 | 0.4 | −0.1 | 0.2 | 0.39 | na |
| KF Moment (%BW·Ht) | **0.2** | **0.4** | **−0.1** | **0.3** | **0.0** | **0.5** | **0.03** | **^0.03** |
| KA Moment (%BW·Ht) | −0.1 | 0.2 | −0.1 | 0.3 | −0.1 | 0.3 | 0.64 | na |
| Hip ROM (°) | 1.0 | 1.9 | 0.7 | 1.2 | 0.9 | 1.6 | | |
| Ankle ROM (°) | 0.5 | 1.7 | 0.1 | 1.7 | 0.3 | 1.7 | | |
| HF Moment (%BW·Ht) | −0.1 | 0.5 | 0.0 | 0.6 | −0.1 | 0.5 | | |
| HE Moment (%BW·Ht) | 0.1 | 0.4 | −0.1 | 0.6 | −0.2 | 0.2 | | |
| ADF Moment (%BW·Ht) | 0.0 | 0.3 | 0.2 | 0.3 | 0.2 | 0.4 | | |

Foundation Doctoral Research Grant (JFH).

## Appendix A. Supplementary data

Supplementary data associated with this article can be found, in the online version, at https://doi.org/10.1016/j.gaitpost.2019.02.008.

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