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A half marathon shifts the mediolateral force distribution at the tibiofemoral joint

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Abstract

Runners' gait patterns vary during a half marathon and influence the knee joint mechanics. Joint contact force is a better estimate of the net joint loadings than external joint moments and closely correlates to injury risks. This study explored the changes of lower limb joint kinematics, muscle activities, and knee joint loading in runners across the running mileages of a half marathon. Fourteen runners completed a half marathon on an instrumented treadmill where motion capture was conducted every 2 km (from 2 km to 20 km). A musculoskeletal model incorporating medial/lateral tibiofemoral compartments was used to process the movement data and report outcome variables at the selected distance checkpoints. Statistics showed no changes in joint angles, muscle co-contraction index, ground reaction force variables, and medial tibiofemoral contact force ($p > 0.05$). Knee adduction moment at 18 km was significantly lower than those at 2 km ($p = 0.002$, $\gamma = 0.813$) and 6 km ($p = 0.001$, $\gamma = 0.663$). Compared to that at 2 km, lateral tibiofemoral contact force was reduced at 18 km ($p = 0.030$, Hedges' $g = 0.690$), 16 km ($p < 0.001$, Hedges' $g = 0.782$), 14 km ($p = 0.045$, Hedges' $g = 0.859$), and 10 km ($p < 0.001$, Hedges' $g = 0.771$) respectively. Mechanical realignment of the lower limb may be the cause of the altered knee loadings and possibly led to reduced running economy in response to a prolonged run. The injury potential of the redistributed tibiofemoral forces warranted further studies.

Keywords: Running; knee joint; computer simulation; biomechanics; sports

Introduction

Marathon has become a popular running event worldwide since its modern renaissance in 1896. A global marathon survey estimated a participation growth of 49.43% from 2008 to 2018 and more than 1.3 million runners accomplishing a full marathon every year.[1] Alongside this booming trend there has been an up to 75% incidence rate of marathon-related injuries.[2] Knee joint is the most vulnerable site to injuries with 8.8% of regular marathoners developing knee osteoarthritis in their careers.[3] Runners are more likely to get injured when running at exhaustion. The physical fatigue state during a prolonged run could affect neuromuscular control, incur faulty movement patterns, and exacerbate injury risks.[4]

Apart from the changes in spatiotemporal parameters (e.g., stride length and cadence), runners were frequently reported to adjust the lower limb kinematics as running mileage amassed in marathon events. These kinematics changes included increased hip internal rotation, reduced knee flexion, and reduced ankle range of motion across running mileages.[5–7] Due to the coupling of segment kinematics and force transmission in the lower limb, these changed joint motions were found to affect shock absorption at landing[8] and influence loads generation at the tibiofemoral articular surface,[9] which in coalescence impose a different injury threat to the knee joint. For example, increased knee adduction moment (KAM) was correlated to higher contact force on the medial knee joint[10] and meniscus pathologies.[11] Previous studies also showed greater workload shear at the knee joint in fatigued runners,[12,13] highlighting the importance of assessing knee joint loadings during a prolonged run and the lack of related evidence in the literature.

Knee adduction moment was often used as a surrogate measure for knee joint loading to predict the severity of symptomatic knee injuries.[14] Though decreased KAMs were demonstrated in various sports maneuvers after a fatigue protocol,[15–17] the timeline of KAM changes in a prolonged run is not clearly documented.[7,18] In addition, KAM is an external measure of the joint loading that can be counteracted by muscle synergy.[14] A fatigue state usually features a weakened stretch-shortening mechanism and force capacity in muscles,[19] which will interact with the external joint moment to influence the internal joint forces. Studying muscle activities and joint contact force would provide more valid measures of the joint loading during running.[14]

Joint contact force determines the bearing stress on the articular surface, which is directly linked to cartilage wear and joint degeneration.[20] It is widely accepted that the tibiofemoral joint force is asymmetrically distributed between the medial and lateral compartments. This uneven force distribution may change over the repeated gaits cycles during a prolonged run and induce injury risk on either side.[21] The recent advancement of musculoskeletal modeling has enabled a separate tibiofemoral contact joint force (TCF) estimation on both medial and lateral components during dynamic tasks.[22] Applying this modeling technique on the half marathon run would help researchers gain a better insight into the loading and injury profile of the sport.

Hence, the study aimed to investigate how accumulated running mileages during a half marathon would change the runners' joint kinematics, muscle activities, and knee joint loading. We calculated the co-contraction index of muscles spanning the medial/lateral knee joint as a means to estimate the muscle activities. The knee joint loading was indicated by KAMs and medial/lateral TCFs. We also presented the results of peak vertical ground reaction force (GRF) and frontal-plane GRF lever arm to the knee to support explanations of the research outcomes. All of the variables are reported at selected distance checkpoints of the half marathon. Based on the previous findings, we hypothesized that, as the half marathon progressed, the runners would increase the hip joint excursion, peak vertical GRF, and GRF lever arm; and reduce the knee flexion, muscle co-contraction, KAM, and TCF.

Materials and methods

Fourteen recreational runners (12 males, 2 females; age = 26.92 ± 6.26 years; height = 1.74 ± 0.07 m; mass = 63.98 ± 8.85 kg) were recruited from the local community. The inclusion criteria were: marathoners with more than two years' race experience, minimally three-race participation each year, no lower limb injuries/symptoms reported at the time of our experiments. A priori sample size calculation for repeated-measures MANOVA was performed upon a pilot study in G*Power (version 3.1.9.7, Heinrich-Heine-Universität Düsseldorf, Düsseldorf, Germany),[23] using the medial/lateral TCFs as the primary dependent variables, the statistical power of $\alpha = 0.05$, $\beta = 0.2$, and an effect size (partial η2) of 0.38 calculated from the data of the first 6 recruited participants of the cohort. The result indicated that a minimum sample size of 12 participants was satisfied. The participants fully understood the research procedure and signed the consent form before study commencement. The study protocol was reviewed and approved by the institutional review board (reference number: HSEARS20170327001).

Before the experiment, 28 retroreflective markers used for motion capture were attached to selected anatomical landmarks based on a published method.[24] The marker attachment was secured with skin-medical tapes (Kinesio Tex Classic, Kinesio, Albuquerque, NM, USA) to avoid markers drop-off during the running trials. The marker trajectories were recorded with a 10-camera motion capture system (T20, Vicon, Oxford Metrics Ltd, Oxford, UK). The participants were given sufficient time to warm up and acclimate themselves to the in-lab running environment. They firstly performed a static standing trial, from which the marker data would be used for model scaling in the subsequent simulation session. After that the participants were asked to use their own running shoes and ran a half marathon on an instrumented treadmill (Bertec Corp., Columbus, Ohio, US) at their preferred speed. The preferred speed for each participant was identified through a testing run before commencements of the actual experiment and kept consistent throughout the entire half-marathon trial.[24] Kinematic and kinetic data were collected for ten consecutive strides at every 2 km of the half marathon, starting at the 2th km and ended at the 20th km.[24] The sampling rates for marker trajectories and GRFs were 250 Hz and 1000 Hz, respectively.

Musculoskeletal modeling was conducted in the Opensim system (version 4.1, National Centre for Simulation in Rehabilitation Research, Stanford, CA, USA).[25] The marker data of the static standing trials were used to scale a generic model[22] to each participant's anthropometries and create a subject-specific model. The generic model was incorporated with a medial-lateral tibiofemoral load distribution algorithm enabled through two independent revolute joints connecting the femoral and tibial segments.[22] The generic model was previously validated and modified based on a standard Opensim model.[25] It featured 22 rigid-body segments, 42 joints, 37 degrees of freedom, and 92 musculotendonous units, which has the robust muscular strength scale to analyze human locomotion dynamics, including walking and running.[25] Following model scaling, inverse kinematics were solved on the data of the running trials to generate the joint coordinates that best reproduced the movement patterns. Marker trajectories and GRFs were filtered using a fourth-order, Butterworth, lowpass filter at 8 Hz and 50 Hz respectively.[24] The mass property of the model and joint coordinates were slightly adjusted based on the results of inverse dynamics to minimize the inconsistency between the predicted and measured external forces. The modulus of forward dynamics in Opensim was then implemented to compute muscle activations and the resultant medial/lateral knee TCFs that drove the model for desired kinematics. KAM was calculated as the external moment resulting from the shank acting on the distal end of the thigh and normalized by body mass and height.[14] Lever arm was defined as the distance of the GRF vector to the knee joint center measured in the frontal plane. The joint kinematics, muscle forces, KAM, and medial/lateral TCFs variables were output in time series by the Opensim Analyze tool.

The degree of muscle co-contraction was quantified by the co-contraction index based on a previous work:[26]

$$CCI_i = \left(F_{1i} + F_{2i}\right) \times \mathrm{MIN}\left(F_{1i}, F_{2i}\right) / \mathrm{MAX}\left(F_{1i}, F_{2i}\right)$$

Where i denotes one data point during the running stance phase. F_{1i} and F_{2i} represent the sums of the forces of two antagonistic muscle groups. MIN and MAX denote the lower and higher values between F_{1i} and F_{2i} respectively. In the study, two co-contraction pairs of muscles were investigated: medial hamstring + medial gastrocnemius head VS. vastus medialis and lateral hamstring + lateral gastrocnemius head VS. vastus lateralis. These two pairs were selected because of their respective contributions to the medial/lateral knee loading.

The primary outcome variables were the maximum medial/lateral TCFs during the stance phase of running gaits. The values of joint kinematics, muscle co-contraction index, GRF, frontal plane knee-GRF lever arm, and KAM were determined at the instant of peak lateral TCF. All of the outcome variables were reported as mean ± standard deviation averaged across ten strides per distance checkpoint per participant and normalized to percentile stance phase.

After log transforms to the data to resume normality, a one-way multivariate analysis of variance (MANOVA) with repeated measures was used to compare the joint kinematics, muscle co-construction indexes, GRF lever arm, and TCFs among the distance checkpoints of the half marathon. A Greenhouse-Geisser correction was applied whenever the assumption of sphericity was not satisfied. Any significant MANOVA results were further investigated by the univariate statistics and post-hoc tests. The p values for pairwise comparisons were corrected by the Benjamini-Hochberg method to control type one errors.[24] Due to the violation of assumptions for parametric tests, the data of KAMs and GRFs were examined by the Friedman test for related samples. For the post-hoc analysis, Wilcoxon signed-rank tests with Bonferroni adjustment were used to identify differences among the distance checkpoints.

The effect sizes were indicated by Hedges' g and γ for MANOVA and Friedman test, respectively, which were interpreted as negligible (< 0.20), small (0.20–0.49), medium (0.50–0.79), and large (≥ 0.80) based on Cohen's convention.[27] All statistics were performed in SPSS software (version 16.0, IBM Corp., Armonk, NY, USA) at the significance level of $\alpha = 0.05$.

Results

The univariate tests (Table 1) following MANOVA ($p < 0.001$) indicated that lateral TCFs were significantly changed by running mileages ($F = 4.245$, $p < 0.001$, partial $\eta 2 = 0.246$) while the joint kinematics, muscle co-contraction index, GRF, lever arm, and medial TCF were not different across the distance checkpoints ($p = 0.102$–0.902). Running mileage also had a significant effect on KAMs (Table 2) as unveiled by the Friedman test ($\chi^2 = 31.153$, $p < 0.001$, Kendall's $W = 0.247$). Post-hoc analyses showed that KAMs at 18 km were significantly lower than those at 2 km ($p = 0.002$, $\gamma = 0.813$) and 6 km ($p = 0.001$, $\gamma = 0.663$). KAMs at 20 km was also lower than that at 6 km ($p = 0.027$, $\gamma = 0.595$). Lateral TCF was found to reduce when the running mileage accumulated from 4 km to 16 km ($p = 0.023$, Hedges' $g = 0.216$) and from 2 km to 10 km ($p < 0.001$, Hedges' $g = 0.771$), 14 km ($p = 0.045$, Hedges' $g = 0.859$), 16 km ($p < 0.001$, Hedges' $g = 0.782$) and 18 km ($p = 0.030$, Hedges' $g = 0.690$) respectively (Figure 1).

Discussion

In this study, we investigated the runners' gait patterns and knee joint mechanics at every 2 km during a half marathon (10 distance checkpoints). The outcomes were expected to provide an insight into the runners' movement strategies in coping with a prolonged run and its implication for joint injuries. We found that the selected variables of joint kinematics, muscle co-contraction index, and medial TCF were not affected by running mileage. In partial accordance with our hypotheses, the runners produced smaller KAMs and lateral TCFs at the later stage of the half marathon compared to the early stage.

Albeit the statistical results, the reduced knee joint loading in our study was not associated with changes in lower limb kinematics, muscle synergy, and GRF. Our observations on the selected running gait features could not generate an explicit explanation for the outcomes of joint forces. Interestingly, previous studies also reported no significant differences in joint angulations and GRFs before and after a similar fatigue running protocol,[7,18,28,29] whereas the effects of a marathon on running gaits may manifest in other biomechanical parameters.[9] The human musculoskeletal system possesses redundant degrees of freedom for a specific locomotion task.[30] Therefore, calculating the single joint angulation may not account for the complexity and nuanced changes in movements that are responsible for the altered knee joint loading. Moreover, it is believed that there is a protective mechanism in the human central nervous system, which could be activated to manage collisions with the ground during fatigued running.[31] In this study, the plausible protective mechanism appeared to function within its adjustment capacity in fine-tuning the lower limb coordination and controlling GRF over the course of a half marathon.[31] The inter-individual variances in running gaits and different adaptation strategies of the marathon runners[29] also diminished the probability of significant statistics.

In the study, we did not measure EMG signals for simulation and validation purposes since a comprehensive model that accounts for muscle fatigue has been lacking. Previous studies consistently found increased EMG signal amplitude in muscles performing repetitive tasks[32] due to the weakened excitation-contraction coupling at exhaustion. In this regard, solving a standard muscle algorithm on fatigue EMG signals would introduce artifacts to the outcomes and most likely overestimate the magnitudes of muscular outputs. Our inverse dynamic approach, on the other hand, may better reflect the load changes under a half marathon condition because the comparisons across distance checkpoints were conducted on comparable baselines by directly resolving the muscle and joint forces on the measured movement data.

A better estimation of actual tibiofemoral joint forces has been the focus of research interests since the increased accessibility of the musculoskeletal modeling technique in the field of human biomechanics. Despite the sparsity of available evidence and variances in model details, our calculations of TCFs were in good agreement with those in the literature. Runners in our study ran at a moderate speed (averaged 3.17 ± 0.43 m/s) and produced the medial and lateral TCFs falling in the range reported for slow walking (medial TCF:1.82 BW, lateral TCF: 1.15 BW) and fast running (medial TCF:5.10 BW, lateral TCF: 2.97 BW).[33–35] Our predicted medial-to-lateral ratios of TCFs (1.44–1.52) were also in accordance with the values in these studies (1.30–1.71), which indicated a reasonable approximation of the load distribution on the tibiofemoral joint.

By the time of this study, there was little evidence concerning the effects of a half marathon run on knee joint loading. Despite the fact, our results of reduced lateral TCFs were in line with findings on medical images. A recent MRI study[36] found that runners who had completed a half marathon showed reduced T2 relaxation time in the lateral femoral region, suggesting that the lateral knee was offloaded after a period of the prolonged run and the subsequent vacuum effects on the articular cavity induced protein expression within the cartilage. We agreed with the authors' contention that the reduced lateral TCFs were caused predominantly by altered alignment of the external force to the lower limb, which could be reflected by the decreased KAMs across the running mileages in our study. A decreased KAM was speculated to alleviate tensile strains within the lateral connective tissues of the knee and decompress the tibiofemoral joint.[37] Interestingly, we found no clear decline of TCFs in the medial compartment. Compared to lateral TCF, medial TCF was more frequently related to external knee moment.[38,39] However, other studies revealed that the magnitude of medial TCF was not only dependent on KAM but also knee moment on the sagittal plane.[38,39] Our results of GRF and its frontal-plane lever arm aggregated and implicated that, the decreased KAM was not achieved through shortening of the moment arm. Instead, the total knee moment might resolve less to KAM and be redistributed on other movement planes as the half-marathon progressed. This assertion was supported by Sanno's findings, which showed increased sagittal-plane knee torques in fatigued runners[13] and explained the insignificant results in medial TCFs.[38]

Within the scope of our observations, it is still unclear why the runners rearranged the alignment of external force to the lower limb and the knee joint loadings during the half marathon. Based on the theory of protective mechanism,[40] mechanical realignment of the lower limb can be accompanied by a concurrent increment of muscle vibration and regulation of muscle stiffness. Though these changes were thought to facilitate attenuating impacts to the body, they may not be favorable to running economy as the knee muscles were considered less efficient in force production than distal leg muscles.[13] Besides, though our statistics did not support increased contact forces on either tibiofemoral compartment, the knee joint loads were clearly showed to redistribute and lean towards the medial side. The force transmitted across the knee seemed sensitive to the subtle adjustment of lower limb alignment.[20] We are concerned that the biomechanical changes presented in the study would be more distinctive and magnified if the running mileage upgrades to a full marathon level. Further investigations for longer running distances (e.g., full marathon and ultra marathon) are warranted in the future.

Some limitations of our study should be acknowledged when interpreting the results. First, the running speed was controlled on the treadmill throughout the half marathon. Though a consistent pacing strategy is advocated by most marathoners, changing running speed is usually inevitable in actual races. Reducing running speed was frequently seen in amateur runners as the concession to fatigue and often incurred different joint biomechanics.[5] The factor was influential to our results but could not be fully addressed in our study. Moreover, the power of our simulation outcomes was confined to some assumptions. Particularly, the musculoskeletal models were unable to consider the fatigued muscle physiologies in the algorithm. Though this technical bottleneck remained to be solved, our results still possess its clinical values from a mechanical standpoint. Finally, the majority of the participants in our study were male. Gender could be a confounding factor because of the differences in anatomical structures, muscle strength, and running biomechanics. Female and male runners may have different abilities to resist exhaustion and respond differently to a prolonged run that requires further investigation.

Conclusion

Runners in our study exhibited no significant changes in lower limb joint angulation, muscle co-contraction, ground reaction force variables, and medial tibiofemoral contact force over the course of a half marathon. They produced lower knee adduction moment and lateral tibiofemoral joint contact force at the later stage of the running trials. In combination, the results suggested that the runners may adapt to the prolonged run by adjusting the mechanical alignment of the lower limb, which explained the redistributed loadings on the knee joint. However, this movement strategy may be carried out at the cost of running efficiency.

Disclosure of interest

The authors report no conflict of interest.

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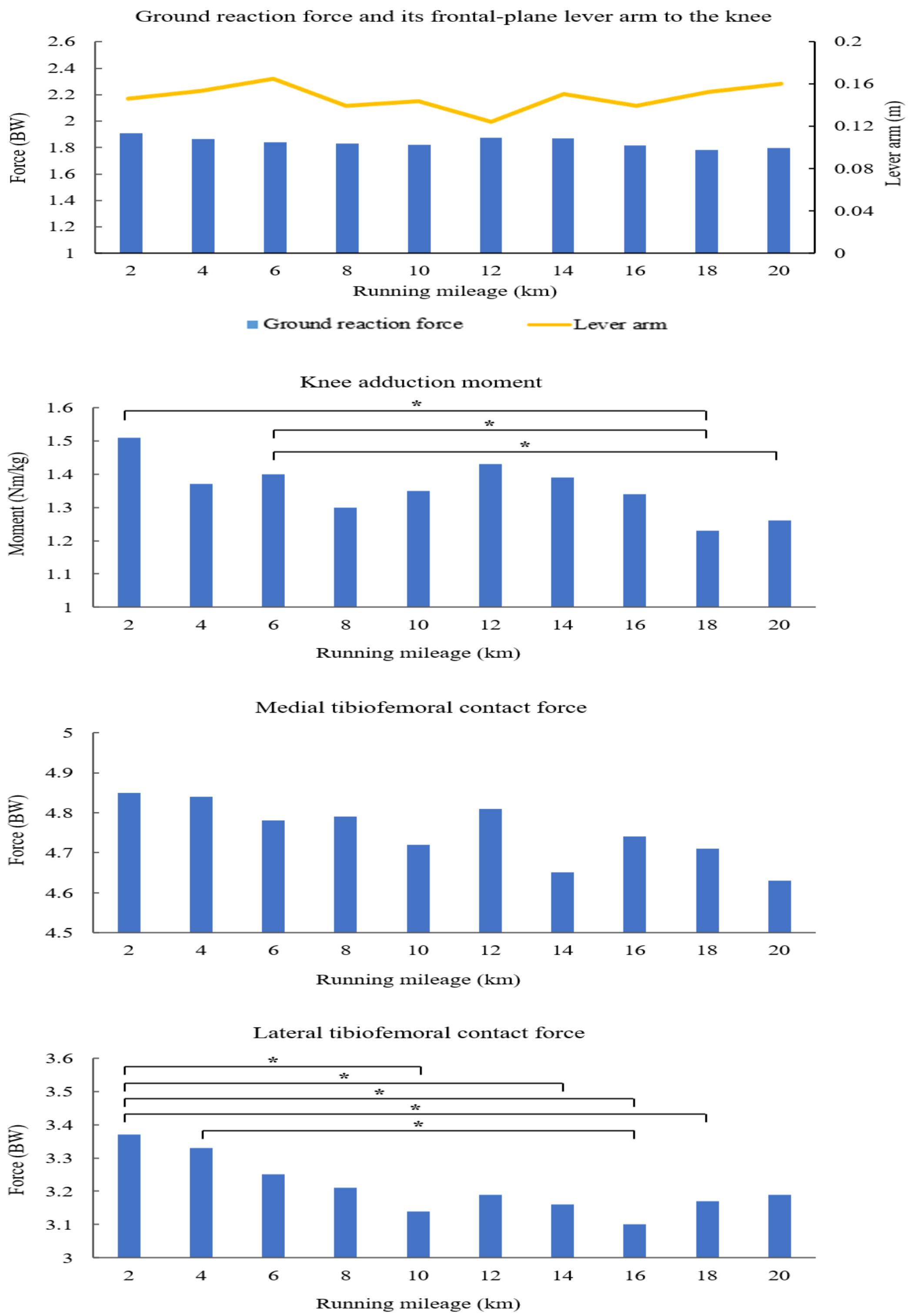
Figure 1. Changes in ground reaction force, frontal-plane ground reaction force lever arm to the knee, knee adduction moment, and medial/lateral tibiofemoral contact force over the course of the half marathon. Acronyms represent: body weight (BW). *$p < 0.05$ in post-hoc pairwise comparison.



Tables

Table 1. The outcome variables across distance checkpoints of a half marathon examined by MANOVA (mean ± standard).

Variables	2 km	4 km	6 km	8 km	10 km	12 km	14 km	16 km	18 km	20 km	F	p	ES (partial η^2)
Joint kinematics (degrees)													
Hip flexion	29.42 ± 5.22	29.58 ± 5.60	30.78 ± 5.38	29.70 ± 6.19	29.42 ± 6.59	28.83 ± 5.42	29.04 ± 6.41	29.43 ± 6.29	29.30 ± 6.21	27.64 ± 5.85	1.581	0.208	0.11
Hip adduction	9.41 ± 4.92	9.37 ± 4.47	9.46 ± 4.75	9.97 ± 4.36	9.45 ± 3.88	10.18 ± 3.93	10.63 ± 4.26	9.92 ± 4.05	9.75 ± 4.32	9.52 ± 4.36	1.677	0.102	0.11
Hip internal rotation	0.59 ± 4.66	0.32 ± 4.37	0.16 ± 4.25	0.18 ± 4.19	0.12 ± 3.57	1.11 ± 3.73	1.41 ± 3.69	0.34 ± 4.03	0.26 ± 3.74	0.27 ± 4.14	1.345	0.278	0.09
Knee flexion	48.18 ± 5.57	48.81 ± 5.46	49.41 ± 5.79	48.91 ± 6.26	47.82 ± 5.52	48.45 ± 5.47	49.13 ± 5.40	48.25 ± 5.94	47.97 ± 5.13	47.78 ± 5.56	1.326	0.276	0.09
Muscle co-contraction index													
Medial knee muscle grounp	1.74 ± 0.77	1.71 ± 0.75	1.74 ± 0.67	1.70 ± 0.79	1.81 ± 0.79	1.79 ± 0.85	1.72 ± 0.94	1.77 ± 0.64	1.85 ± 0.64	1.74 ± 0.69	0.300	0.902	0.02
Lateral	0.88	0.86	0.87	0.86	0.91	0.89	0.82	0.94	0.92	0.90	0.59	0.66	0.04

knee muscle group	± 0.41	± 0.42	± 0.36	± 0.45	± 0.42	± 0.44	± 0.46	± 0.50	± 0.40	± 0.51	0	1	
Frontal-plane ground reaction force lever arm to the knee (m)													
Lever arm	0.14 ± 0.12	0.15 ± 0.11	0.16 ± 0.10	0.14 ± 0.08	0.15 ± 0.12	0.13 ± 0.10	0.15 ± 0.10	0.14 ± 0.11	0.15 ± 0.09	0.16 ± 0.09	0.938	0.495	0.07
Knee joint contact force (BW)													
Medial compartment	4.85 ± 0.60	4.84 ± 0.55	4.78 ± 0.49	4.79 ± 0.50	4.72 ± 0.47	4.81 ± 0.51	4.65 ± 0.47	4.74 ± 0.48	4.71 ± 0.47	4.63 ± 0.39	1.826	0.179	0.12
Lateral compartment	3.37 ± 0.59	3.33 ± 0.66	3.25 ± 0.55	3.21 ± 0.54	3.14 ± 0.46	3.19 ± 0.50	3.16 ± 0.45	3.10 ± 0.49	3.17 ± 0.57	3.19 ± 0.52	4.245	< **0.001**	0.24

Statisitical significant differences are bold. Acronyms represent: body weight (BW), effect size (ES).

Table 2. The outcome variables across distance checkpoints of a half marathon examined by Friedman test (mean ± standard deviation).

Variables	2 km	4 km	6 km	8 km	10 km	12 km	14 km	16 km	18 km	20 km	X^2	p	ES (Kendall's W)
Knee	1.5	1.3	1.4	1.3	1.3	1.4	1.3	1.3	1.2	1.2	31.15	<	0.25

adduction moment (Nm/kg)	1 ± 0.43	7 ± 0.40	0 ± 0.42	0 ± 0.34	5 ± 0.39	3 ± 0.34	9 ± 0.39	4 ± 0.30	3 ± 0.35	6 ± 0.34	3	**0.001**	
Vertical ground reaction force (BW)	1.91 ± 0.29	1.86 ± 0.21	1.84 ± 0.23	1.83 ± 0.20	1.82 ± 0.24	1.87 ± 0.28	1.87 ± 0.21	1.81 ± 0.19	1.79 ± 0.25	1.80 ± 0.15	5.688	0.771	0.05

Statisitical significant differences are bold. Acronyms represent: body weight (BW), effect size (ES).