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Ankle kinetics and plantarflexor morphology in older runners with different lifetime running exposures

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ABSTRACT

Running promotes better cardiovascular health and has positive effects on the musculoskeletal system in older adults. However, older adults have lower ankle plantarflexor torques and positive powers during running, and exhibit changes in plantarflexor morphology than young adults. Since older runners who run as much as younger runners exhibit youthful ankle mechanical outputs, running exposure may preserve the locomotor factors that mediate running speed. The purpose of this study was to compare ankle mechanical output during running and plantarflexor morphological characteristics between older runners who have low or high lifetime running exposure. Ten older runners with low lifetime running exposure and nine older runners with high lifetime running exposure performed over-ground running trials at 3.0 m/s (±5%) while kinematic and ground reaction force (GRF) data were collected and used to compute joint angular kinetics. Right medial gastrocnemius morphological characteristics were assessed using ultrasonography at rest and during isometric contractions. Ankle torques, powers, and plantarflexor morphology were compared between groups. Older runners with different lifetime running exposures ran with similar ankle mechanical output (i.e. no effect of running exposure) ($p > .05$) and exhibited similar medial gastrocnemius morphology during isometric testing. The findings from this study demonstrate that lifetime running exposure does not appear to influence ankle mechanical output or plantarflexor morphology in middle-aged runners.

1. Introduction

The number of adults over 40 years participating in running races has been steadily increasing with a 21% increase in participation between 1980 and 2013 and with ~19% of U.S. marathon participants being over the age of 55 years in 2013 (Lamppa, 2013). Considering the declining health status of North Americans over recent decades, increased running participation could mitigate risks of death from all causes of cardiovascular disease (Lee et al., 2014), lower body mass index (P. T. Williams, 2013), and training is generally associated with better overall physical performance regardless of age (Tanaka & Seals, 2008; Tarpenning, Hamilton-Wessler, Wiswell, & Hawkins, 2004). However, many physiological and biomechanical factors are negatively affected by aging and may ultimately reduce movement performance and keep aging adults from participating in exercise such as running. It is therefore

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important to better understand what factors can influence the physiology and biomechanics of aging runners to lessen and/or delay the onset of age-related declines in function.

Aging is associated with reduction in muscle volume (Morse, Thom, Birch, & Narici, 2005) total number of muscle fibers and cross-sectional area (Lexell, Taylor, & Sjöström, 1988), and reduction in fiber fascicle length of the gastrocnemius (Mademli & Arampatzis, 2008; Stenroth, Peltonen, Cronin, Sipilä, & Finni, 2012). These age-related changes in muscle characteristics can in part be attributed to more sedentary behavior with aging (Evenson, Buchner, & Morland, 2012; Harvey, Chastin, & Skelton, 2013), as a reduction of weight bearing activity is associated with a decrease in fascicle length and pennation angle of gastrocnemius (Reeves, Maganaris, Ferretti, & Narici, 2002). These declines in plantarflexor complex characteristics may explain why older adults generate lower plantarflexion torques compared to younger adults (Thom, Morse, Birch, & Narici, 2007). Since mechanical and morphological properties of the gastrocnemius can influence their capacity to generate torque at the ankle (Thom et al., 2007), older runners have a reduced ability to generate ankle plantarflexor angular power and work compared to younger runners (DeVita et al., 2016; Fukuchi, Stefanyshyn, Stirling, Duarte, & Ferber, 2014). These age-related reductions in ankle kinetics are paralleled with lower late stance peak vertical ground-reaction force (GRF) (Kline & Williams III, 2015), anterior propulsive GRF (Korhonen et al., 2009), and step length (DeVita et al., 2016) in older compared to younger runners. Consequently, aging leads to slower preferred (DeVita et al., 2016) and maximal (Korhonen et al., 2009) running speeds. Slower running speeds during training may be in part responsible for age-related declines in maximal and submaximal cardiorespiratory function, strength, and power (Quinn, Manley, Aziz, Padham, & MacKenzie, 2011). Ultimately, maintenance of ankle joint kinetics and plantarflexor morphology and mechanical properties are perhaps essential in maintaining function during activities of daily living (Stenroth et al., 2015).

Muscle architecture and tendon mechanical properties have an effect on muscle fiber length and velocity and as a result, can influence muscle force production and power generation according to the force-length relationship (Gordon, Huxley, & Julian, 1966) and force-velocity relationship (Edman, 1988). For example, at the same muscle shortening velocity, a muscle with longer fibers can produce greater force and power compared to muscles with shorter fibers due to lower sarcomere velocity. Further, muscle fiber pennation angle may alter the amount of force applied to a tendon by altering the angle of pull relative to the force-generating axis. A smaller pennation angle can transmit more of the muscles force to its corresponding tendon as opposed to a larger pennation angle (Lieber & Fridén, 2001). Thus, it is reasonable to think that longer muscle fibers have smaller pennation angles and therefore, are desired for more optimal force transmission to its tendon. Cross-sectional studies have shown that endurance running does not significantly alter fascicle length despite larger observed pennation angles in endurance runners compared to untrained controls, regardless of age (Abe, Kumagai, & Brechue, 2000; Karamanidis & Arampatzis, 2005). However, previous studies have typically assessed the influence of running exposure without accounting for *lifetime* running exposure. More lifetime running exposure potentially leads to positive morphological changes that may be related to the preservation of certain joint kinetics that are negatively affected by aging. In fact when weekly running volume is matched, older runners run with similar peak positive ankle power than young runners (Paquette, DeVita, & Williams III, 2018) and, highly-trained older runners exhibit similar leg and vertical stiffness found in younger similarly trained runners (Pantoja, Morin, Peyré-Tartaruga, & Brisswalter, 2016). Although lifetime running exposure does not appear to influence running mechanics in distance runners between the age of 15 and 54 years (Agresta, Peacock, Housner, Zernicke, & Zendler, 2018), it is currently unclear how lifetime exposure may influence ankle kinetics and plantarflexor morphological characteristics in runners over 50 years. This information would help delineate the influence of training on running gait function to understand if training exposure contributes to continued running participation with aging.

The purpose of this study was to compare ankle kinetics during running and medial gastrocnemius morphology properties in older runners with different lifetime running exposures. We hypothesized that older runners with higher lifetime running exposure would run with longer steps, and greater propulsive GRF, peak ankle plantarflexor torques, peak ankle positive power, and ankle positive work compared to runners with lower lifetime running exposure. We also hypothesized that older runners with higher lifetime running exposure would have larger pennation angles, shorter fascicles, and greater stiffness of the medial gastrocnemius compared to runners with lower lifetime running exposure.

2. Materials and methods

2.1. Participants

Nineteen runners (9 women and 10 men) over the age of 50 years were recruited for this study (Table 1). Participants were included if they had no lower extremity injuries within the past 12 months, if they ran at least 10 miles per week, completed at least three runs per week, and if they had no metabolic or orthopedic conditions. Prior to participation, each participant was informed of all procedures, potential risks, and benefits associated with the study through both verbal and written form in accordance with the procedures approved by the University Institutional Review Board for Human Participants Research.

2.2. Procedures

All participants filled out a questionnaire regarding their training experience and other training-related details. Participants were separated into low (LOW) or high (HIGH) lifetime training exposure. Lifetime training exposure was defined as the product of number of lifetime running years, average weekly lifetime running mileage, and average number of weeks running per year (Agresta et

Table 1
Participant and training characteristics for both LOW and HIGH groups (means ± SD).

Characteristics	Low	High	*p*-value	*d*
Age (years)	60 ± 7	62 ± 6	0.62	0.34
Height (m)	1.73 ± 0.10	1.68 ± 0.10	0.22	0.53
Mass (kg)	73.0 ± 9.4	67.0 ± 14.6	0.57	0.52
Body Mass Index ($kg \cdot m^{-2}$)	24.5 ± 3.0	23.6 ± 3.6	0.37	0.30
Years of lifetime Running **	10.6 ± 3.1	36.7 ± 8.4	0.00	4.45
Weekly Volume over lifetime ($miles \cdot week^{-1}$) *	50.6 ± 3.1	49.3 ± 3.2	0.35	0.46
Weekly Volume in Past Years ($miles \cdot week^{-1}$) *	18.1 ± 3.7	25.7 ± 5.8	0.005	1.68
Years without running	0.3 ± 0.6	2.2 ± 3.1	0.08	0.94
Lifetime Running Exposure (miles) **	9456 ± 2934	45,983. ± 14,967	0.00	3.68
Preferred Running Pace ($min \cdot mile^{-1}$)	11.0 ± 1.5	10.1 ± 2.1	0.11	0.57

Notes: *: $p < .05$, **$p < .001$; LOW: low running exposure; HIGH: high running exposure; *d*: Cohen's effect size.

al., 2018). Participants also filled out the Foot and Ankle Outcome Score survey to assess ankle joint health: a possible confounding factor in our analysis (Martin & Irrgang, 2007; Roos, Brandsson, & Karlsson, 2001).

Since either the forefoot or heel is the first point of contact with the ground during running, differences in foot structure and characteristics may result in differences in mechanics of the entire lower extremity (Nigg, Cole, & Nachbauer, 1993; Powell, Williams 3rd, Windsor, Butler, & Zhang, 2014; Williams III, Davis, Scholz, Hamill, & Buchanan, 2004). The Arch Height Index Measurement System was used to obtain right foot and ankle measurements during sitting and standing to enable the computation of the Arch Height Index (D. S. Williams & McClay, 2000) since high arched runners have been shown to have smaller peak knee abduction moments when compared to low-arched runners (Powell, Andrews, Stickley, & Williams, 2016).

To establish a target tensile force during testing, the Arch Height Index Measurement System was used to obtain foot and ankle measurements of the right limb while standing. To control for perspective error examiner recorded measurements from Arch Height Index Measurement System by orienting line of vision to be directly perpendicular to the scales of the measurement system. All measurements were obtained by the same examiner for all participants included in this study. Achilles tendon moment arm (MA_{AT}) was measured as the antero-posterior distance between the lateral malleolus and the posterior aspect of the Achilles tendon over the skin (Fig. 1). The external moment arm (MA_{EXT}) was measured as the distance between the point of force application (i.e. first metatarsal) and lateral malleolus (Fig. 1).

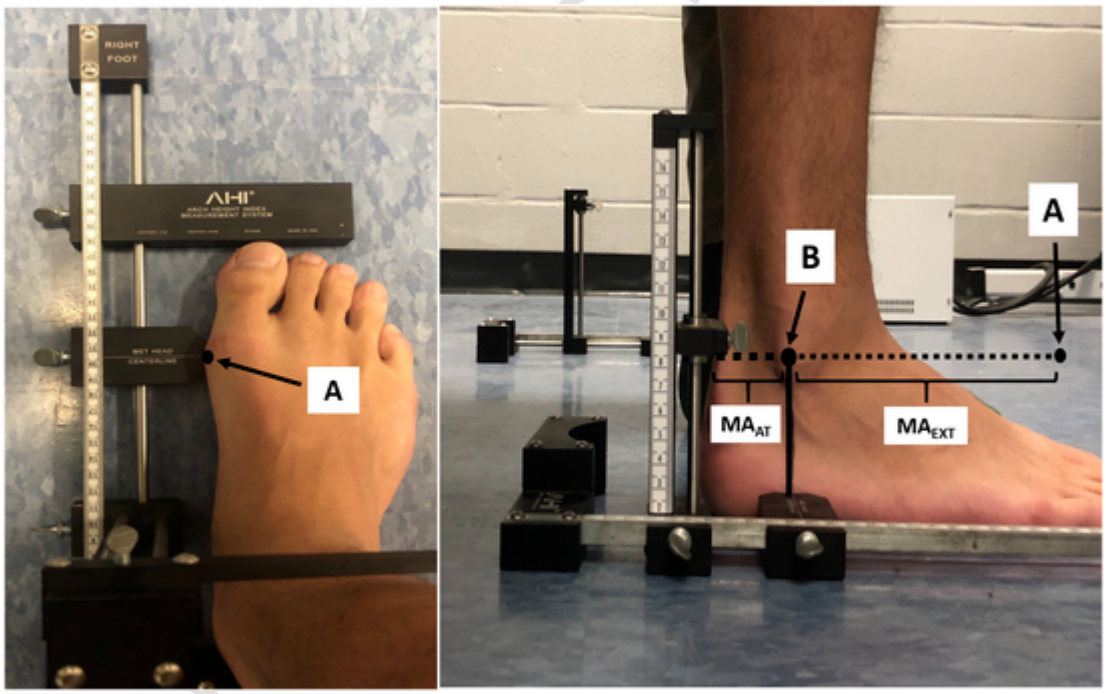

Fig. 1. Achilles tendon moment arm (MA_{AT}) and external moment arm (MA_{EXT}) where measured using the Arch Height Index Measurement System. A) represents the point of the first metatarsal head; which was taken form ARHI measurements prior to this estimation. B) Represents the position of the lateral malleolus, which was estimated using a vertical line drawn using a 90-degree angle that was placed flush with the floor prior to foot placement in the Arch Height Index Measurement System.

Participants then completed a five-minute running warm-up at their preferred running speed on a treadmill (Excite+ RUN NOW, TechnoGym, USA) but all experimental testing took place over-ground. All participants wore standard lab shoes (New Balance, MX623) during running tests. A 9-camera three-dimensional (3D) motion capture system (240 Hz, Qualisys AB, Göteburg, Sweden) and a 3D force platform (1200 Hz, AMTI, Watertown, MA, USA) were used to obtain 3D kinematics and ground reaction forces (GRFs), respectively. These data were collected synchronously using Qualisys Track Manager Software (Qualisys AB, Gotenburg, Sweden). Data were collected on the right leg of each participant using 12.7 mm spherical reflective markers placed on anatomical landmarks to define each segment. Clusters of four non-collinear markers secured to thermoplastic shells were used to track pelvis, thigh, and shank motion. The pelvis was defined using the iliac crests and greater trochanters, and the hip joint center was calculated at the location of one-quarter the distance between ipsilateral and contralateral greater trochanter (Weinhandl & O'Connor, 2010). The thigh was defined with the greater trochanter, the calculated hip joint center, and femoral epicondyles. The shank was defined with the femoral epicondyles and the malleoli. Additionally, the foot was defined with the malleoli and the first and fifth metatarsal heads. A thermoplastic shell with three reflective markers was placed on the heel of the right shoe to track the segment. Finally, one reflective marker was placed on the heel of the left shoe to allow step length measurements. A one-second static calibration trial was recorded before the start of data collection to define joint centers, segment coordinate systems, and segment dimensions. Anatomical markers were removed, and participants performed five running trials over a 25 m runway at a set speed of 3.0 m/s (±5%). This speed was monitored using an electronic timer (54035A, Lafayette Instruments Inc., IN, USA) and two photocells (63,501 IR, Lafayette Instruments Inc., IN, USA) placed three meters apart at shoulder height for all participants on tripods perpendicular to the laboratory runway. Two to three practice trials were provided to allow the participants to achieve the desired testing speed while contacting the force platform with their right foot without visual targeting.

Following the running trials, assessments of medial gastrocnemius (MG) morphology and mechanical characteristics were performed. A force transducer (Model MLP-1 k, Transducer Techniques, Temecula, CA) attached in-series with a metal chain was secured underneath a treatment table (Fig. 2). On the other end, the metal chain was attached to a cuff secured to a wooden platform at a location in line with the 5th metatarsal head to stabilize the foot during testing (Fig. 2). Muscle morphological characteristics of MG were assessed using a linear array ultrasound transducer (L12-4 MHz Philips, Lumify; USA), positioned along the longitudinal axis of the right MG at 30% of the distance between the malleoli and the femoral epicondyles (Pamukoff & Blackburn, 2015) which is approximately 50% of muscle length (Kubo et al., 2003). The ultrasound probe was secured to the leg using Velcro straps attached to a custom-designed plastic mold (Autodesk, Inventor Pro) and fabricated with a 3D printer (Makerbot 5th Generation Replicators, USA). With this arrangement, the only force acting on the probe was gravity, thereby reducing variations in application of pressure over the skin. Participants were secured in the prone position to a treatment table using a harness system (Solo-Step System, USA) to avoid any longitudinal movements of the body during contractions (Fig. 2). Further, the harness system tension could be manipulated to move the participant in a position to ensure a 90-degree ankle joint angle during contractions (Figs. 2 & 3). Before experimental testing, participants were asked to gradually push against the wooden platform while the chain and harness straps were under maximal tension to ensure that the ankle joint remained stable and within three degrees of the initial 90-degree position during testing. For MG stiffness estimates, muscle fascicle length was measured at different target tensile forces. Specifically, these target tensile forces were equivalent to 25%, 50%, 75%, and 100% of previously reported plantarflexor (PF) forces during running

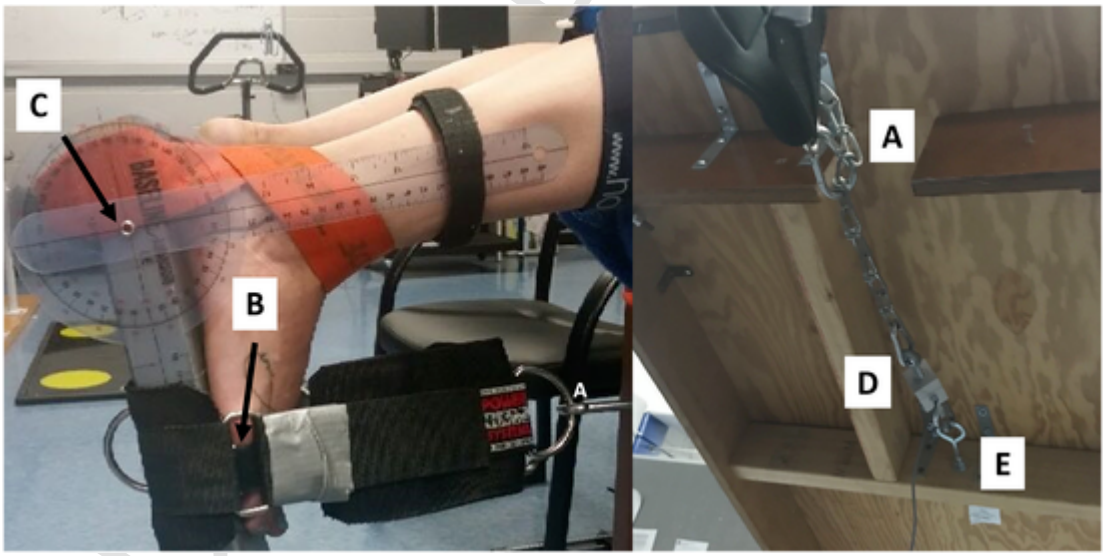

Fig. 2. Foot placement on the wooden platform attached to the metal chain in-series with the strain gauge. A) Represents the point of chain attachment with the cuff; B) represents the mid-foot (5th metatarsal head) alignment with the cable attached to the force transducer; C) goniometer position used to ensure that the ankle remained within the three-degree position window; D) point of chain attachment to the table, and; E) force transducer attached to chain and table to enable force measurement during contractions.

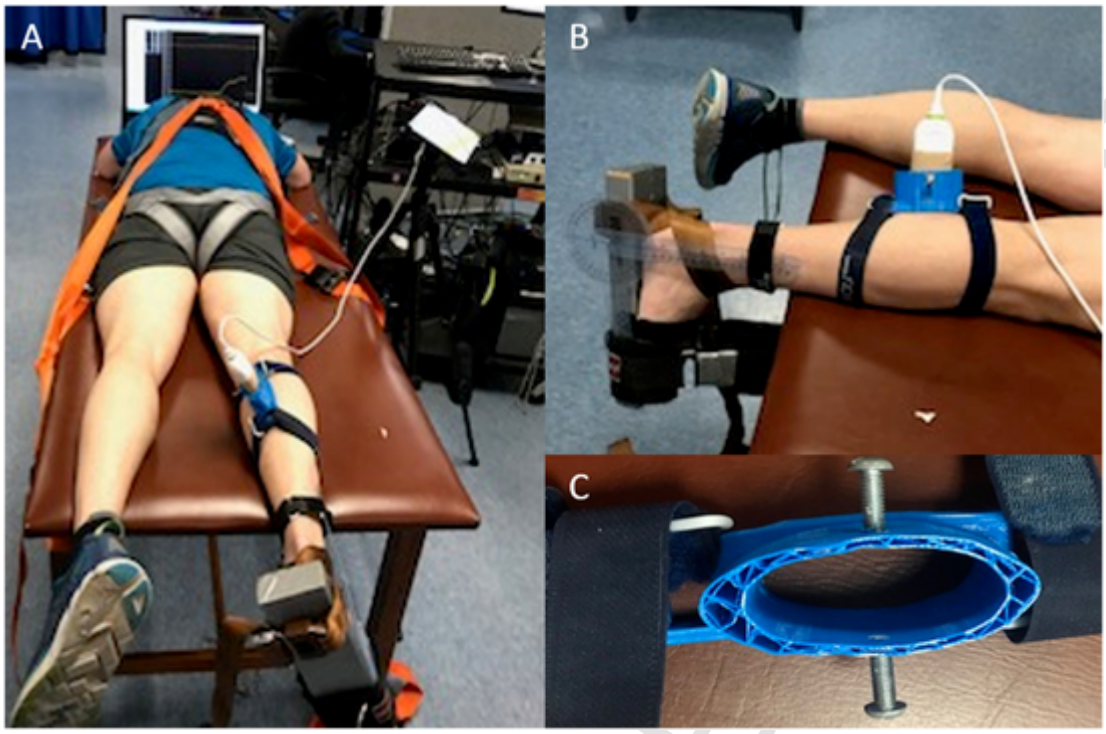

Fig. 3. Medial gastrocnemius morphological assessment experimental set up. A) The harness system and safety straps that were used to position and secure the participant to the testing table; B) the ultrasound probe positioning and orientation, and; C) custom-designed plastic mold used to position the probe over the skin above the medial gastrocnemius.

(Besier, Fredericson, Gold, Beaupré, & Delp, 2009). To obtain individual participant estimated target tensile force, previously reported peak gastrocnemius force (i.e., sum of medial and lateral gastrocnemius) force during running (~24.15 $N \cdot kg^{-1}$) was multiplied by each participant's body mass. The force transducer target tensile force during testing was calculated from each participant's Achilles tendon moment arm (MA_{AT}) and external moment arm (MA_{EXT}) (location of force application below the 5th metatarsal) measured earlier in the testing session, and the estimated PF force (Eq. 1).

$$Tensile\ Force = \frac{\left(PF\ force\ x\ MA_{AT}\right)}{MA_{EXT}} \tag{2}$$

During testing, participants were asked to produce the 25%, 50%, 75%, and 100% target tensile forces streamed in real-time with Qualisys Track Manager Software on a computer screen that was positioned in their line of vision. Participants were asked to gradually increase their force output by pushing against the wooden foot platform until the target tensile force threshold was reached and to hold for approximately two to three seconds. Two images per target tensile force were captured separately at rest and during isometric contractions for each tensile force percentage. Rest periods of approximately 30 s was provided between contractions, while ultrasound image quality was verified. If the quality of the images was deemed to be of poor quality (i.e., blurry), an additional image was taken to ensure clear differentiation between the deep aponeurosis and FL.

All ultrasound images and measurements were completed by the same examiner to ensure good intra-rater reliability. Intra-rater reliability of the ultrasound measurements in the population included in the study was not assessed. However, reliability of the morphological measures taken 24 h apart by the same examiner was assessed during pilot testing within our laboratory using a young adult population. Reliability of pennation angle was good with intra-class correlation coefficients (Koo & Li, 2016) ranging between 0.61 and 0.93.

2.3. Data analyses

Visual3D (C-Motion, Germantown, MD) was used to compute 3D joint kinematic and kinetic variables. Kinematic data were interpolated using a least-squared fit of 3rd order polynomial with a three-data point fitting and maximum gap of 10 frames. Kinematic and GRF data were filtered using a fourth-order Butterworth low-pass filer with cut-off frequencies of 8 and 40 Hz, respectively. A right-hand rule with a Cardan rotational sequence (x-y-z) was used for 3D angular computations, where x represents the sagittal plane, y represents the frontal plane, and z represents the transverse plane. A vertical GRF threshold of 20 N was used to define the start and end of the stance phase. The ankle joint angular kinematic and kinetic variables were expressed in the shank coor-

dinate system. Dependent running biomechanics variables included: step length; peak propulsive GRF; peak ankle plantarflexor torque; peak positive ankle angular power, and ankle joint angular work. Newtonian inverse dynamics was used to compute net internal ankle joint torque ($Nm \cdot kg^{-1}$) during stance. The 3D angular ankle joint powers ($W \cdot kg^{-1}$) were computed as the dot product of joint torque and angular velocities. Step length was computed as the anterior-posterior distance between the left heel marker at time of left foot contact and a right heel marker at time of right foot contact. Since only one force plate was used under the right foot, a kinematic-based method using the peak downward velocity of the pelvis center of mass was used to identify time of left foot contact (Milner & Paquette, 2015). Custom software (Matlab 2017, Mathworks, Natick, MA, USA) was used to calculate ankle angular work as angular power integrated with respect to time using the trapezoidal rule. The average of each dependent variable from the five running trials was used in statistical analyses.

The two ultrasound images obtained from each contraction condition were analyzed separately using an open-source program (Image J 1.44b, National Institutes of Health). MG fascicle pennation angles (θ) were measured as the angle of insertion of muscle fascicle into the deep aponeurosis (Fig. 4). Fascicle length (FL) was calculated using a previously reported equation (Kubo et al., 2003) (Eq. 2; Fig. 4). Each measurement was taken three times for each of the two images per contraction condition and the average of the six measurements was used to compute the change in FL (ΔFL) between FL rested and FL contracted (Eq. 3) for each target tensile force.

$$FL\ (mm) = \frac{MT(mm)}{Sin\ \theta} \tag{3}$$

$$\Delta FL\ (mm) = FL\ rested - FL\ contracted \tag{4}$$

Due to tension on the strain gauge and inevitable movement of the ankle joint during the rested condition, the initial force reading from the force-time curve was variable and often fluctuated between measurements. To control for this fluctuation, force values at rest were subtracted from their respective target tensile forces to establish a "zero" baseline prior to each contraction at all four percentages. Finally, MG tensile forces were calculated as 70% (i.e., the percent contribution of MG relative to total gastrocnemius peak force (Dorn, Schache, & Pandy, 2012) of the measured tensile forces at each contraction percentage and were plotted against the MG ΔFL from the resting and contracting images and MG stiffness was calculated as the slope of the force and ΔFL curve for each participant.

2.4. Statistical analyses

Independent *t*-tests were used to compare biomechanical variables and plantarflexor morphology and characteristics, and participant and training characteristics between groups. Data normality was assessed using the Kolmogorov-Smirnov tests ($p < .05$). If data were not normally distributed a Mann-Whitney non-parametric test was used to compare group differences. Significance was set at an alpha level of 0.05. Cohen's *d* effect sizes were also calculated for effect magnitude of group mean differences (i.e., small: $d \leq 0.6$, moderate: $0.6 > d < 1.2$; large: $d \geq 1.2$ (Hopkins, 2011).

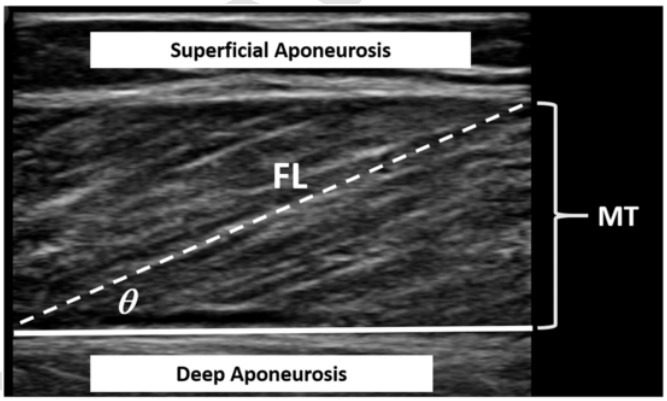

Fig. 4. Image measurements of the medial gastrocnemius muscle. NOTE: Fascicle Length (FL) was calculated using the following eq. FL = MT x Sin θ^{-1}: where θ is the pennation angles, and MT is muscle thickness.

3. Results

3.1. Participant characteristics

We confirmed that the HIGH group had greater lifetime running exposures, lifetime years of running and, average weekly miles in the past year compared to the LOW group (Table 1). Additionally, Foot and Ankle Outcome Score and Arch Height Index measures were not statistically different between groups (Table 5) confirming similar ankle joint health and foot types between groups.

3.2. Running biomechanics

No group differences were observed for step length, peak propulsive force, and ankle plantarflexor torque, peak positive power, and positive work (Table 2).

3.3. Medial gastrocnemius morphology

Morphological measurements of MG pennation angle and estimation of FL for all resting and contracted conditions were not different between running exposure groups (Table 3). Estimated MG forces during testing were not different between groups (Table 4) which was expected considering they are directly related to body mass and no group differences were observed in body mass (Table 1). Additionally, ΔFL were not different between groups at any of the rested and contracted conditions (Table 4). Finally, MG stiffness was not different between groups (Table 4).

Table 2
Step length, peak propulsive force, and ankle joint kinetics for both LOW and HIGH groups (mean ± SD).

Variables	Low	High	*p*-value	*d*
Step Length (m)	0.98 ± 0.08	0.96 ± 0.07	0.68	0.28
Peak Propulsive Force (BW)	0.25 ± 0.09	0.19 ± 0.06	0.11	0.82
Peak Ankle Plantarflexor Torque ($Nm \cdot kg^{-1}$)	−2.57 ± 0.64	−2.16 ± 0.83	0.10	0.62
Peak ankle positive power ($Nm \cdot kg^{-1}$)	8.19 ± 2.41	6.7 ± 2.43	0.20	0.66
Negative Ankle Work ($J \cdot kg^{-1}$)	−0.41 ± 0.10	−0.45 ± 0.12	0.07	0.42
Positive Ankle Work ($J \cdot kg^{-1}$)	0.64 ± 0.20	0.51 ± 0.21	0.15	0.69

Notes: *: $p < .05$; LOW: low running exposure; HIGH: high running exposure. Step length: anterior distance between right heel position at time of right foot contact and left heel position at time of left foot contact; *d*: Cohen's effect size.

Table 3
Morphological measurements of medial gastrocnemius at rest and contracted for both LOW and HIGH groups (mean ± SD).

Variables	Low	High	*p*-value	*d*
25% Contraction				
Θ_R (°)	18.9 ± 2.4	18.0 ± 1.3	0.36	0.49
Θ_C (°) *	21.2 ± 2.9	19.6 ± 2.0	0.21	0.66
FL_R (mm)	59.1 ± 14.1	52.7 ± 7.8	0.57	0.58
FL_C (mm)	51.9 ± 13.8	47.5 ± 6.9	0.42	0.42
50% Contraction				
Θ_R (°) *	18.9 ± 2.4	15.8 ± 5.5	0.19	0.79
Θ_C (°) *	22.4 ± 4.2	20.6 ± 1.3	0.26	0.58
FL_R (mm)	58.8 ± 14.1	53.8 ± 8.7	0.39	0.45
FL_C (mm)	48.7 ± 15.4	44.8 ± 6.3	0.51	0.34
75% Contraction				
Θ_R (°) *	19.3 ± 2.8	17.5 ± 1.4	0.13	0.83
Θ_C (°) *	23.5 ± 4.4	21.5 ± 1.6	0.23	0.64
FL_R (mm)	57.6 ± 13.7	54.0 ± 7.9	0.52	0.34
FL_C (mm)	46.3 ± 14.7	43.4 ± 6.17	0.61	0.27
100% Contraction				
Θ_R (°) *	19.0 ± 2.6	17.8 ± 1.4	0.23	0.63
Θ_C (°) *	25.2 ± 5.3	22.5 ± 2.2	0.19	0.70
FL_R (mm)	58.4 ± 14.1	53.6 ± 7.7	0.40	0.44
FL_C (mm)	44.4 ± 14.7	42.1 ± 6.6	0.67	0.21

Notes: *: $p < .05$; LOW: low running exposure; HIGH: high running exposure. Θ_R: pennation angle at rest; Θ_C: pennation angle during contractions; FL_R: muscle fascicle length at rest; FL_C: muscle fascicle length during contractions; *d*: Cohen's effect size.

Table 4
Change in muscle fascicle length (ΔFL) between rested and contracted conditions, estimated medial gastrocnemius (MG) forces in contracted conditions, and MG stiffness for both LOW and HIGH groups (mean ± SD).

Variables	Low	High	*p*-value	*d*
ΔFL at 25% (mm)	7.2 ± 2.6	5.2 ± 2.3	0.12	0.83
ΔFL at 50% (mm)	10.1 ± 3.8	8.9 ± 3.2	0.49	0.36
ΔFL at 75% (mm)	11.3 ± 3.4	10.6 ± 2.6	0.63	0.26
ΔFL at 100% (mm)	14.2 ± 4.9	11.5 ± 2.8	0.20	0.68
MG Force at 25% (N)	142.0 ± 31.2	132.0 ± 7.2	0.39	0.45
MG Force at 50% (N)	260.4 ± 51.0	231.1 ± 26.4	0.16	3.63
MG Force at 75% (N)	379.2 ± 71.0	330.8 ± 38.3	0.10	0.88
MG Force at 100% (N)	498.5 ± 922.0	432.8 ± 52.4	0.09	0.10
MG Stiffness ($N{\cdot}mm^{-1}$)	37.1 ± 21.8	39.8 ± 21.4	0.80	0.13

Notes: LOW: low running exposure; HIGH: high running exposure; *d*: Cohen's effect size.

Table 5
Foot and Ankle Outcome Scores for each category out of 100 and Arch Height Index measured for both LOW and HIGH groups (mean ± SD).

Variables	Low	High	*p*-value	*d*
Pain	98.1 ± 3.5	99.1 ± 1.3	0.88	0.40
Symptoms	96.8 ± 5.2	97.6 ± 4.5	0.96	0.18
Activities of Daily living	99.4 ± 1.0	99.8 ± 0.5	0.30	0.57
Sports & Recreation	99.5 ± 1.5	99.4 ± 1.6	0.94	0.04
Quality of Life	98.1 ± 4.0	93.8 ± 8.84	0.23	0.69
Arch Height Index	0.33 ± 0.02	0.32 ± 0.01	0.20	0.66

Notes: *: $p < .05$; LOW: low running exposure; HIGH: high running exposure. *d*: Cohen's effect size.

4. Discussion

The purpose of this study was to compare ankle joint kinetics during running and medial gastrocnemius morphological properties in older runners with different lifetime running exposures. Considering the popularity of running as a mode of aerobic exercise and the increased running participation by older adults, understanding the cumulative lifetime effects of running is important.

We hypothesized that older runners with higher lifetime running exposure would run with longer steps, greater propulsive GRF, larger peak plantarflexor torque, peak positive ankle power, and ankle positive work compared to runners with lower lifetime exposure. Contrary to this hypothesis, ankle kinetics, step length, and peak propulsive GRF were not different between lifetime running exposure groups. This suggests that higher lifetime running exposure in older runners might not help preserve ankle joint mechanical output and propulsive function during running. These findings are similar to recent evidence that lifetime running exposure in *young to middle-aged* runners does not influence spatio-temporal and joint kinematics and GRF variables during running (15–54 years) (Agresta et al., 2018). It therefore seems that running volume exposure has limited effects on ankle motion and kinetics and GRF variables in runners between approximately 15 and 70 years. Further, although other studies did not report lifetime running exposure of their middle-aged or older runners, the peak plantarflexor torque (~2.2–2.6 $Nm{\cdot}kg^{-1}$) and peak positive ankle power (~6.7–8.2 $Nm{\cdot}kg^{-1}$) of both groups in the current study are similar to previously reported ankle torques (~2.1–2.3 $Nm{\cdot}kg^{-1}$) and powers (~7.3–7.6 $W{\cdot}kg^{-1}$) in middle-aged and older runners (DeVita et al., 2016; Paquette et al., 2018). These values suggest that older runners in our study ran with ankle kinetics that are lower than magnitudes typically reported in young runners (DeVita et al., 2016; Kuhman, Melcher, & Paquette, 2016; M. R. Paquette, Zhang, & Baumgartner, 2013; Sinclair, 2014). It has been recently hypothesized that preferred running pace (i.e., intensity) may have a greater influence on ankle mechanical output than average weekly running volume in middle-aged runners (Paquette et al., 2018). Additionally, older sprint-trained adults tend to have a greater Achilles tendon cross-sectional area compared to older endurance trained and inactive adults (Stenroth et al., 2015). Thus, our current findings suggest that higher lifetime running exposure measured using weekly mileage in runners over 50 years may not be a large enough stimulus to alter ankle mechanical output during running. Perhaps a better lifetime training exposure measure would include cumulative average intensity of running, or pace, throughout a lifetime and future studies should investigate this hypothesis.

Since age-related changes in morphological properties of MG (i.e., shorter fascicles) may contribute to enhanced resistance to muscle fatigue during submaximal sustained isometric contraction (Mademli & Arampatzis, 2008), we expected that lifetime exposure to a task that regularly require fatiguing exercise bouts, such as running, may result in a similar adaptation due to the demands of the activity. Thus, we also hypothesized that runners with higher *lifetime running exposure* would have larger pennation angles, shorter fascicle length, and greater stiffness of the medial gastrocnemius when compared to runners with low lifetime running exposure. Our hypothesis was rejected as no differences in pennation angle, fascicle length and stiffness estimations were observed be-

tween groups. These observations are consistent with previous findings that no differences in MG fascicle length is observed in older runners compared to older non-active individuals (Karamanidis & Arampatzis, 2005, 2006; Stenroth, Cronin, et al., 2015). Our results also contradict previous findings that MG pennation angles are greater in older runners compared to older non-active individuals (Karamanidis & Arampatzis, 2005, 2006). Furthermore, MG pennation angles and fascicle lengths are not significantly different among older inactive adults, older endurance-trained, or older sprint-trained adults (Stenroth, Cronin, et al., 2015). In the current study we assessed MG morphological characteristics but did not assess the characteristics of soleus, lateral gastrocnemius, and the Achilles tendon which contribute to ankle mechanical output during locomotion (Karamanidis & Arampatzis, 2006, 2007). Taken together these findings suggest that, even when accounting for lifetime exposure, MG morphological characteristics assessed isometrically are not different between older non-runners and runners, and between middle-aged runners with less or more running exposure.

This study has several limitations. First, the cross-sectional design of the current study makes it impossible to truly understand the longitudinal effects of lifetime running exposure, especially since participants recall of training history may not have been completely accurate. With the cross-sectional designs of the current and previous studies, lifetime running training effects may only be inferred. It is therefore evident that longitudinal studies are needed to understand the influence of lifetime training exposures on morphological properties of muscles. Secondly, target tensile forces used during MG morphological testing for each participant were estimated from previously reported average peak gastrocnemius forces (i.e. sum of MG and LG) during over-ground running (Besier et al., 2009) and the participants body mass. Therefore, these estimations of MG force do not represent the individual forces of each runner in the current study. Thirdly, these target tensile forces did not include the force contribution from soleus during running and therefore, underestimating actual peak plantarflexor forces given the large contributions of soleus during running (Dorn et al., 2012). However, there is evidence that soleus cross-sectional area and pennation angle are not different between older runners and older inactive controls suggesting that soleus peak force might not be influenced by running volume in older runners (Stenroth, Cronin, et al., 2015). Therefore, it might be suggested that the omission of soleus would not influence the estimated MG stiffness between the LOW and HIGH lifetime volume groups in the current study. Additionally, it is possible that slightly larger ankle motion could have resulted from the isometric contractions. However, our methods were performed meticulously to limit ankle angle during plantarflexor isometric testing to a maximum of three degrees as explained in the methods section. Our estimate of fascicle length is highly dependent on the pennation angle measurements and a direct measurement of the fascicle should be considered in future studies. Finally, the average preferred pace was 11.0 and 10.1 min·mile^{-1} (or 2.4 and 2.7 m·s^{-1}) for the LOW and HIGH groups, respectively. Thus, the testing speed of 3.0 m·s^{-1} might have required greater relative effort and lower limb joint mechanical demands for the LOW vs the HIGH group during testing. However, given the short distance covered during testing (i.e., 25 m) and the rest in between trials, the slightly faster testing speed relative to self-reported preferred pace in both groups likely did not influence the findings of this study.

5. Conclusions

Findings from this study suggest that ankle propulsive function deficits might not be affected by higher lifetime running exposure, and that lifetime running exposure does not appear to influence morphological and mechanical properties of medial gastrocnemius in middle-aged runners. Therefore, it appears that running exposure may be insufficient to attenuate the functional propulsive deficits observed in aging runners. More specific interventions focused on improving plantarflexors and Achilles Tendon capacity, such as training modalities with loading time greater than 3 s and heavy resistance exceeding 80% of maximal voluntary contraction (Bohm, Mersmann, & Arampatzis, 2015; Rønnestad & Mujika, 2014; Schoenfeld, Ogborn, & Krieger, 2017), might be more beneficial than just running training to combat age-related declines in propulsive function and performance. Also, since middle-aged runners who have similar preferred running paces as younger runners run with the same ankle mechanical output (Paquette et al., 2018), cumulative average intensity of running or pace throughout a lifetime might be a better measure of lifetime training exposure in aging runners. Continued research on this topic may delineate if the rate of age-related declines in plantarflexor function can be moderated.

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CRediT authorship contribution statement

Ramzi M. Majaj: Conceptualization, Formal analysis, Investigation, Validation, Methodology, Visualization, Writing - original draft, Writing - review & editing. **Douglas W. Powell:** Conceptualization, Investigation, Methodology, Writing - original draft, Writing - review & editing. **Lawrence W. Weiss:** Conceptualization, Investigation, Methodology, Writing - original draft, Writing - review & editing. **Max R. Paquette:** Conceptualization, Formal analysis, Investigation, Methodology, Visualization, Writing - original draft, Writing - review & editing, Supervision.

Declaration of Competing Interest

The authors have no conflicts of interest to declare.

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