

Effects of a Fatigue Protocol on Vertical, Leg, and Joint Stiffness during Overground Running

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Abstract

Running-induced neuromuscular fatigue can alter spring-mass characteristics such as vertical, leg, and joint stiffness. The purpose of the study was to investigate the effects of a treadmill run fatigue protocol on stiffness parameters during overground running. Sixteen healthy cadets (22.4 ± 3.9 yr, 1.76 ± 0.08 m, 72.4 ± 12.3 kg) from a University's Army Reserve Officer Training Corp were included in the study. Rested and exhausted state overground running biomechanics were collected prior to and immediately after a fatigue protocol that involved an Åstrand Protocol Graded Exercise Test and an exhaustive run at a velocity associated with 80% $\text{VO}_{2\text{max}}$. No significant changes were observed for vertical, leg, or joint stiffness group means. While insignificant, knee stiffness displayed increasing trends while hip stiffness displayed decreasing trends. Leg length at initial contact (-1%, $p = 0.007$) and compressed leg length (-1%, $p = 0.013$) significantly decreased. Hip excursion (+9%, $p = 0.021$), change in knee moment (+7%, $p = 0.027$), and knee moment at the instance when the anterior-posterior ground reaction force was zero (+8%, $p = 0.021$) significantly increased. While running in a fatigued state, subjects attempted to maintain overall vertical and leg stiffness with joint stiffness modulations.

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Introduction

Neuromuscular fatigue can occur from repeated muscle stretch-shortening cycles during moderate to high-intensity running and can be quantified by increased ratings of perceived exertion (RPE)¹⁻⁴ or reductions in performance characteristics such as maximum voluntary contraction force,⁴⁻⁶ and jump and landing performance.^{1,6,7} Neuromuscular fatigue also affects spring-mass characteristics such as vertical (K_{vert}), leg (K_{leg}), and joint stiffness (K_{joint}), as well as ground contact time, flight time, step frequency and length, and vertical ground reaction force during running.^{1-3,8-22}

Running has been described by a spring-mass model where components of the lower limb act as a spring to absorb ground reaction forces (GRF) and propel the center of mass (COM) during ground contact. Stiffness of the spring is related to the aforementioned spring-mass characteristics and has been shown to be inversely related to oxygen cost of running at a given velocity.^{23,24} Runners who use a combination of short ground contact time, high step frequency, and long flight time display a higher K_{vert} or K_{leg} , and produce greater running economy, that is decreased oxygen consumption at a given velocity, than runners with more compliant leg spring characteristics.²³⁻²⁶ Decreased K_{vert} and K_{leg} is associated with reduced running economy as a result of a more compliant leg spring that is inefficient at storing potential elastic energy in the musculotendinous units and the inability to convert it to kinetic energy for propulsion during the stretch shortening cycle.^{23,24}

Modulation of stiffness can be theorized as using either a more bony absorption strategy with increased stiffness, or a more muscular and soft tissue absorption strategy with decreased stiffness.²⁷ As muscular fatigue inevitably occurs during prolonged running, increasing stiffness would be ideal to reduce the load on the musculature. However, research has shown that effects

of fatigue on *Kvert* or *Kleg* can vary.^{2,9,28} During steady-state running in a non-fatigued state, reported *Kvert* values have been 20.55-58.95 kN/m.^{1-3,8,10,17,18,20,21,28,29} Due to fatigue, *Kvert* has increased 2.19-4.2 kN/m,^{2,3,17} decreased by 6-8.7%^{9,20} and 2.54-6.09 kN/m,^{15,16} or did not change.^{2,3,20,21,28,29} Leg stiffness values during steady-state running in a non-fatigued state have been reported as 6.16-15.5 kN/m.^{8,10,11,16-18,20-22,28,29} Similar to *Kvert*, running induced fatigue has caused *Kleg* to increase 1.3 kN/m,¹⁷ decrease 0.8-1.84 kN/m,^{10,16,20,21} or not change.^{11,18,22,28,29}

Research investigating *Kjoint* during fatigued running is limited compared to *Kvert* and *Kleg*. Luo et al.¹⁵ observed *Kankle* decreased, *Kknee* increased, and *Khip* trended towards a significant increase ($p = 0.07$). Although there were more net increases to *Kjoint* of the lower limb, *Kvert* decreased at a level approaching significance ($p = 0.06$) due to greater COM displacement.¹⁵ In agreement to Luo et al.,¹⁵ Weir et al.²² found *Kknee* increased and *Kankle* decreased, while *Kleg* remained the same during two prolonged runs under different shoe conditions. The inverse *Kjoint* change at the ankle and knee likely influenced the lack of change to *Kleg*.²² In contrast, a previous study by Radzak et al.,³ which utilized the same treadmill running fatigue protocol as the current study, observed *Kknee* decreased despite no changes to *Kvert*. This could suggest *Kknee* was decreased to accommodate an increased *Kankle* or *Khip* to modulate *Kvert*. However, *Kjoint* of the ankle and hip, and *Kleg*, was not included in the study for further description of stiffness changes of the entire leg spring. Since *Kleg* involves the angle of the lower limb during contact, it is possible the hip excursion component of *Khip* can have a certain degree of contribution to *Kleg*. However, research investigating the contribution of *Khip* to *Kleg* is limited. Further investigation of how individual *Kjoint*'s contribute to *Kleg* is needed to understand how the body modulates spring-mass behavior in response to fatigue.

Therefore, the purpose of the study was to investigate the effects of a treadmill running fatigue protocol on *Kvert*, *Kleg*, and *Kjoint* of the ankle, knee, and hip during overground running. Additionally, the study aimed to describe the relationship between *Kleg* and *Kjoint* of the ankle, knee, and hip. Based on the effects of similar intensity and duration fatigue protocols on *Kvert* and *Kleg*,^{2,3,10,13,20,21} we hypothesized that *Kvert* would not change and *Kleg* would decrease. Due to the current study utilizing the same fatigue protocol as Radzak et al.³, we hypothesized that *Kknee* would decrease. In line with two studies that investigated fatigue effects on *Kankle*^{15,22} and one study that investigated hip *Kjoint*,¹⁵ we hypothesized that *Kankle* would decrease and *Khip* would not change.

Methods

Subjects

Sixteen healthy cadets (mean $\pm$ SD: age = 22.4 ± 3.9 years, height = 1.76 ± 0.08 meters, mass = 72.4 ± 12.3 kilograms) from the University of Hawaii at Manoa's Army Reserve Officer Training Corp who were part of a larger project¹ were included in the current study. Subjects were healthy with no activity restrictions and were considered as low risk according to the ACSM Risk Stratification Categories.³⁰ Subjects completed an informed consent form approved by the University Institutional Review Board and a brief medical history questionnaire which was evaluated by a Certified Athletic Trainer. Subjects wore their own non-standardized running shoes during data collection. All subjects reported right limb dominance.

Treadmill Run Protocol

A speed-blinded treadmill run protocol was conducted on a Quinton Medtrack T65 Treadmill (Cardiac Science, Corp. Bothell, WA). A metabolic cart containing an Oxygen Analyzer and Carbon Dioxide Analyzer (AEI Technologies, Naperville, IL) was used to collect metabolic data via open circuit indirect calorimetry. Ambient temperature, barometric pressure, and relative humidity were recorded from a Davis VantageVUE (Davis Instruments, Hayward, CA, USA). Prior to each data collection session, the metabolic cart was calibrated according to manufacturer specifications. Subjects were fitted with a head support and mouthpiece with a two-way non-rebreather valve which was connected to the metabolic cart. A nose clamp was fitted to ensure breathing was isolated to the mouthpiece. A Polar Pacer T31 heart rate monitor (Polar Electro Oy, Finland) was fitted on the sternum to collect heart rate.

Subjects underwent a Modified Åstrand Protocol Graded Exercise Test (GXT) to determine VO₂max.³¹ Standardized instructions were stated prior to the GXT, which emphasized the importance of maximal effort by the participant. Running speed for the GXT was determined via the participants' perceived comfortable thirty-minute running pace, between 5-8 mph. Subjects were blinded to the treadmill speed throughout the testing procedure. The GXT was terminated when subjects reached volitional exhaustion and VO₂max was confirmed based upon meeting one of the following criterion: a respiratory exchange ratio greater than 1.15, a RPE $\geq$ 17, or a plateau in maximal oxygen output with an increase in work rate.³¹ Following the GXT, the breathing apparatus was removed and subjects were given a three-minute, self-selected pace walking recovery at a 1% grade. The breathing apparatus was then refitted, and the exhaustive run began. Treadmill speed was increased to a speed predicted to elicit 80% VO₂max at 1% grade as determined by the American College of Sports Medicine (ACSM) equations for estimating oxygen consumption.³¹ For the first three minutes of the exhaustive run, metabolic data were collected to determine if adjustments to treadmill speed were necessary to elicit 80% of VO₂max. The breathing apparatus was removed following verification of 80% $\pm$ 5% VO₂max intensity at 1% grade. Subjects continued running at the prescribed treadmill speed while the grade increased 2.5% every three minutes until volitional exhaustion. Differentiated RPE³² were collected following the completion of the GXT, prior to the start of the exhaustive run, and following the exhaustive run. Immediately after the exhaustive run, retroreflective markers were replaced as necessary for fatigued state running gait trials.

Running Biomechanics Data Collection

Running biomechanics were collected prior to and immediately after the treadmill fatigue protocol. Preceding rested state running biomechanics data collection, subjects were allowed a

self-directed warm-up and familiarization running trials on the 18-m runway to ensure constant running velocity ($4.0 \text{ m/s} \pm 10\%$). Speedtrap II (Brower Timing Systems, Draper, UT) infrared sensors were placed four meters apart on the middle third of the runway to collect running velocity. Twenty-seven retroreflective markers were placed on the subjects' lower limbs and thorax according to a modified Plug-in-Gait model. A static calibration trial was conducted to determine joint centers with the use of medial anatomical markers. After the static calibration trial, medial anatomical markers were removed. Three-dimensional running kinematics were captured at 240 Hz using a Vicon motion capture system with 13 cameras and a Vicon Nexus software (Vicon, inc., Centennial, CO) to capture, reduce, and analyze kinematic data. Kinetic data were collected at 960 Hz with an AMTI force platform (Advanced Medical Technology Incorporated, Boston, MA) embedded flush within the 18-m runway and time synchronized with kinematic data. Three successful trials per limb were recorded, and mean values for dominant right limb were used for running gait analysis. During the rested state, subjects were allowed to walk back to the start position following each trial. Subjects' rating of perceived exertion (RPE) according to Borg's 6-20 scale were collected prior to, at midpoint, and after completion of running biomechanics data collection.

Data Processing

Running biomechanics data of the right limb were filtered using a fourth-order, low-pass Butterworth filter with a 10 Hz cutoff frequency for kinematics and 50 Hz cutoff frequency for kinetics. Joint moments were calculated using inverse dynamics and reported as external moments. Joint flexion angles and external flexion/dorsiflexion moments were reported as positive values.

Vertical stiffness (K_{vert}) was calculated with equation (1)²³:

$$K_{vert} = F_{max} / \Delta y \quad (1)$$

where F_{max} was the maximum vertical GRF of the stance phase and Δy was the COM vertical displacement from initial contact to the lowest position achieved during stance.

Leg stiffness (K_{leg}) was calculated with equation (2)³³:

$$K_{leg} = F_{max} / \Delta L \quad (2)$$

where F_{max} was the maximum vertical GRF and ΔL was the change in vertical leg length.

Change in vertical leg length was calculated using equation (3)³³:

$$\Delta L = \Delta y + L_0(1 - \cos \theta_L) \quad (3)$$

where Δy was COM vertical displacement and L_0 was leg length at initial contact. θ_L was the half angle of the arc swept by the leg and was calculated with equation (4)³³:

$$\theta_L = \sin^{-1}(ut_c / 2L_0) \quad (4)$$

where u was the forward velocity of COM during stance and t_c was ground contact time.

Joint stiffness (K_{joint}) was calculated with equation (5)³⁴:

$$K_{joint} = \Delta M / \Delta \theta \quad (5)$$

where ΔM was the change in external joint moment and $\Delta \theta$ was the joint excursion. ΔM was calculated with equation (6):

$$\Delta M = M_{AP\ Zero} - M_{IC} \quad (6)$$

where $M_{AP\ Zero}$ was the external joint moment at the instance when the anterior-posterior ground reaction force vector was zero (AP Zero) and M_{IC} was the external joint moment at initial contact. $\Delta \theta$ was calculated with equation (7):

$$\Delta \theta = \theta_{AP\ Zero} - \theta_{IC} \quad (7)$$

where $\theta_{AP\ Zero}$ was the joint angle at AP Zero and θ_{IC} was the joint angle at initial contact.

Statistical Analysis

Statistical analyses were performed with SPSS 26.0 with an alpha priori level set at $p \leq 0.05$. Multiple repeated measures analysis of variance (ANOVA) was used to compare rested state and fatigued state running variables. Partial eta squared (η_p^2) was used to estimate effect size (small $\eta_p^2 = 0.01$, medium $\eta_p^2 = 0.06$, large $\eta_p^2 = 0.14$). Descriptive statistics are presented as mean value $\pm$ standard deviation. Percent change of variables from rested state to fatigued state were used to descriptively analyze trends with clinical relevance defined as greater than $\pm 5\%$ change.

Results

All subjects met at least one of the criteria for VO₂max attainment during the exhaustive protocol. Graded exercise test duration was $8:50 \pm 1:46$ min and VO₂max was 51.35 ± 9.51 mL/kg/min. The duration of the exhaustive run at $80\% \pm 5\%$ VO₂max was $17:55 \pm 3:37$ min at 3.00 ± 0.56 m/s treadmill speed. Mean time from exhaustive protocol completion to onset of fatigue state running gait trials was $5:53 \pm 1:03$ min.

Kvert and *Kleg* group means did not change significantly from rested to fatigued state. Lack of significant changes to *Kvert* and *Kleg* were reflected by no significant changes to *Fmax*, Δy , and ΔL . Significant decreases were observed for *L_o* ($p = 0.007$) and compressed leg length (*L_c*) ($p = 0.013$). The other ΔL components (*u*, *t_c*, and θ_L) were not significantly different between rested and fatigued state running gait trials. Means and standard deviations for rested state and fatigue state *Kvert*, *Kleg*, and related components are presented in Table 1.

No significant changes were observed for *Kankle*, *Kknee*, or *Khip* group means. $\Delta\theta_{hip}$ significantly increased +9% ($p = 0.021$), ΔM_{knee} significantly increased +7% ($p = 0.027$), and *M_{knee AP Zero}* significantly increased +8% ($p = 0.021$). Means and standard deviations for rested state and fatigue state *Kjoints* and related components are presented in Table 2.

Table 1. Repeated measures ANOVA results for *Kvert*, *Kleg*, and related parameters.

	Rested			Fatigued			$\Delta\%$	p	η_p^2
Vertical Stiffness Parameters									
K_{vert} (kN/m)	30.41	±	6.52	30.23	±	5.84	-1%	0.766	0.01
K_{vert} (kN/kg/m)	0.42	±	0.05	0.42	±	0.06	0%	0.982	0.00
F_{max} (kN)	1.84	±	0.32	1.86	±	0.31	+1%	0.319	0.07 ^{&}
F_{max} (kN/kg)	25.41	±	1.93	25.76	±	2.56	+1%	0.235	0.09 ^{&}
Δy (m)	0.062	±	0.007	0.063	±	0.007	+2%	0.443	0.04
Leg Stiffness Parameters									
K_{leg} (kN/m)	10.61	±	2.41	10.41	±	2.25	-2%	0.365	0.06 ^{&}
K_{leg} (kN/kg/m)	0.15	±	0.02	0.14	±	0.02	-2%	0.402	0.05
ΔL (m)	0.176	±	0.021	0.181	±	0.020	+3%	0.107	0.16 [#]
L_0 (m)	0.932	±	0.049	0.925	±	0.050	-1%	0.007*	0.39 [#]
L_c (m)	0.817	±	0.052	0.806	±	0.052	-1%	0.013*	0.35 [#]
θ (°)	25.99	±	1.97	26.55	±	1.80	+2%	0.167	0.12 ^{&}
t_c (s)	0.22	±	0.02	0.22	±	0.02	0%	0.284	0.08 ^{&}
u (m/s)	4.06	±	0.17	4.07	±	0.19	< +1%	0.820	0.00

Mean ± sd, percent change, p value, and partial eta squared for vertical stiffness (*Kvert*), (*Kleg*), and related variables (maximum vertical GRF (*Fmax*), vertical COM displacement (Δy), vertical leg length change (ΔL), leg length at initial contact (L_0), compressed leg length (L_c), half angle of the arc swept by the leg (θ_L), ground contact time (t_c), and forward velocity of COM (u)) during rested state and fatigued state.

* Fatigued state significantly different from rested state ($p \leq 0.05$)

Large effect size ($\eta_p^2 = 0.14$)

& Medium effect size ($\eta_p^2 = 0.06$)

Table 2. Repeated measures ANOVA for *Kjoint* and related parameters.

	Rested			Fatigued			$\Delta\%$	p	η_p^2
Ankle Stiffness Parameters									
<i>Kankle</i> (Nm/°)	10.36	±	4.23	11.62	±	3.43	+12%	0.308	0.07 ^{&}
<i>Kankle</i> (Nm/kg/°)	0.15	±	0.06	0.16	±	0.05	+12%	0.325	0.06 ^{&}
$\theta_{ankle\ IC}$ (°)	4.23	±	10.62	6.19	±	8.66	+46%	0.176	0.12 ^{&}
$\theta_{ankle\ AP\ Zero}$ (°)	24.89	±	4.31	25.78	±	5.05	+4%	0.207	0.10 ^{&}
$\Delta\theta_{ankle}$ (°)	-20.67	±	9.14	-19.59	±	7.14	+5%	0.356	0.06 ^{&}
$M_{ankle\ IC}$ (Nm/kg)	-0.026	±	0.030	-0.037	±	0.028	-41%	0.076	0.20 [#]
$M_{ankle\ AP\ Zero}$ (Nm/kg)	2.85	±	0.46	2.82	±	0.41	-1%	0.594	0.02
ΔM_{ankle} (Nm/kg)	2.88	±	0.46	2.86	±	0.41	-1%	0.730	0.01
Knee Stiffness Parameters									
<i>Kknee</i> (Nm/°)	8.27	±	3.44	8.69	±	2.74	+5%	0.264	0.08 ^{&}
<i>Kknee</i> (Nm/kg/°)	0.11	±	0.03	0.12	±	0.03	+7%	0.139	0.14 [#]
$\theta_{knee\ IC}$ (°)	14.45	±	7.08	16.89	±	5.92	+17%	0.072	0.20 [#]
$\theta_{knee\ AP\ Zero}$ (°)	39.68	±	5.62	41.53	±	7.00	+5%	0.070	0.20 [#]
$\Delta\theta_{knee}$ (°)	25.23	±	6.22	24.64	±	5.07	-2%	0.540	0.03
$M_{knee\ IC}$ (Nm/kg)	-0.31	±	0.15	-0.32	±	0.16	+3%	0.549	0.02
$M_{knee\ AP\ Zero}$ (Nm/kg)	2.32	±	0.43	2.50	±	0.47	+8%	0.021*	0.31 [#]
ΔM_{knee} (Nm/kg)	2.64	±	0.47	2.82	±	0.55	+7%	0.027*	0.29 [#]
Hip Stiffness Parameters									
<i>Khip</i> (Nm/°)	6.88	±	4.81	5.84	±	2.78	-15%	0.267	0.08 ^{&}
<i>Khip</i> (Nm/kg/°)	0.09	±	0.05	0.08	±	0.04	-15%	0.207	0.10 ^{&}
$\theta_{hip\ IC}$ (°)	42.26	±	6.85	44.20	±	7.86	+5%	0.141	0.14 [#]
$\theta_{hip\ AP\ Zero}$ (°)	27.11	±	6.47	27.70	±	7.74	+2%	0.644	0.02
$\Delta\theta_{hip}$ (°)	15.15	±	4.57	16.50	±	4.81	+9%	0.021*	0.31 [#]
$M_{hip\ IC}$ (Nm/kg)	0.78	±	0.36	0.75	±	0.32	-4%	0.731	0.01
$M_{hip\ AP\ Zero}$ (Nm/kg)	-0.52	±	0.51	-0.51	±	0.52	+2%	0.910	0.00
ΔM_{hip} (Nm/kg)	1.31	±	0.69	1.27	±	0.62	-3%	0.734	0.01

Mean $\pm$ sd, percent change, p value, and partial eta squared for ankle stiffness (*Kankle*), knee stiffness (*Kknee*), hip stiffness (*Khip*), joint angles at initial contact (θ_{IC}), joint angles at AP Zero ($\theta_{AP\ Zero}$), joint excursions ($\Delta\theta$), joint moments at initial contact (M_{IC}), joint moments at AP Zero ($M_{AP\ Zero}$) and change in joint moments (ΔM).

* Fatigued state significantly different from rested state ($p \leq 0.05$)

Large effect size ($\eta_p^2 = 0.14$)

& Medium effect size ($\eta_p^2 = 0.06$)

While statistically insignificant, clinically relevant trends ($> \pm 5\%$) were observed. Three outliers skewed *Kjoint* group mean percent changes. Subject AROTC-240 increased *Khip* +1018% with a +888% increased ΔM_{hip} , subject AROTC-249 increased *Khip* +110% with a +94% increased ΔM_{hip} , and subject AROTC-253 increased *Kankle* +1022% with a +432% increased $\Delta \theta_{ankle}$. Excluding the three outliers, an inverse change between *Kknee* (+6%) and *Khip* (-20%) group means were observed, whereas *Kankle* did not have a clinically relevant change (+1%). No clinically relevant changes were observed for *Kvert* and *Kleg* group means, including and excluding the three outliers. Dorsiflexion $\theta_{ankle IC}$ (+46%) and plantarflexion $M_{ankle IC}$ (+41%) group means at initial contact increased proportionate to each other. Group mean $\theta_{knee IC}$ increased +17% with no clinically relevant change to $\Delta \theta_{knee}$. Within this sample, 50% of subjects experienced a decrease in *Khip*, 56% experienced an increase in *Kknee*, 43% experienced an increase in *Kankle*, and 37% experienced a decrease in *Kankle*. 37% decreased *Kvert*, 25% increased *Kvert*, and 37% did not change *Kvert*. 31% increased *Kleg*, 37% decreased *Kleg*, and 31% saw no change to *Kleg*. Table 3 presents individual subjects' raw *Kvert* and *Kleg* values during rested state, fatigued state, and the change between both conditions. Table 4 presents individual subjects' raw *Kjoint* values during rested state, fatigued state, and the change between both conditions. Figure 1 presents individual subjects' and group mean *Kjoint* percent changes and Figure 2 presents *Kvert* and *Kleg* percent changes, excluding the three outliers.

Table 3. Subject *Kvert* and *Kleg* values.

Subjects	<i>Kvert</i> (kN/m)			<i>Kleg</i> (kN/m)		
	Rested	Fatigued	Δ	Rested	Fatigued	Δ
AROTC-204	32.36	34.51	+2.15	10.57	11.37	+0.80
AROTC-230	28.29	26.32	-1.97	8.17	7.95	-0.22
AROTC-238	26.48	23.47	-3.01	7.94	6.86	-1.08
AROTC-239	26.38	28.12	+1.74	8.48	8.53	+0.05
AROTC-240	39.35	32.99	-6.36	11.25	10.06	-1.19
AROTC-242	30.72	31.49	+0.77	10.92	10.47	-0.45
AROTC-248	32.21	31.60	-0.61	11.79	12.05	+0.26
AROTC-249	32.98	32.28	-0.70	12.33	12.57	+0.24
AROTC-251	30.38	28.39	-1.99	10.61	11.22	+0.61
AROTC-252	48.38	47.34	-1.04	17.12	15.91	-1.21
AROTC-253	27.56	27.21	-0.35	8.82	9.94	+1.12
AROTC-257	21.30	22.61	+1.31	8.12	7.21	-0.91
AROTC-261	31.65	31.79	+0.14	13.45	12.17	-1.28
AROTC-263	24.88	28.62	+3.74	9.69	10.43	+0.74
AROTC-264	31.03	32.75	+1.72	11.65	10.32	-1.33
AROTC-266	22.63	24.16	+1.53	8.92	9.53	+0.61

Individual subject's vertical stiffness (*Kvert*) and leg stiffness (*Kleg*) raw values during rested and fatigued state, and the change between the two conditions.

Table 4. Subject *Kjoint* values.

Subjects	<i>Kankle</i> (Nm/deg)			<i>Kknee</i> (Nm/deg)			<i>Khip</i> (Nm/deg)		
	Rested	Fatigued	Δ	Rested	Fatigued	Δ	Rested	Fatigued	Δ
AROTC-204	17.56	13.61	-3.95	6.39	7.66	+1.27	7.76	7.72	-0.04
AROTC-230	9.93	16.39	+6.46	6.85	10.06	+3.21	5.86	5.38	-0.48
AROTC-238	11.22	10.7	-0.52	6.76	7.23	+0.47	2.29	2.18	-0.11
AROTC-239	14.99	11.44	-3.55	6.66	8.33	+1.67	5.05	3.28	-1.77
AROTC-240	9.84	11.55	+1.71	15.73	14.08	-1.65	0.38	4.22	+3.84
AROTC-242	7.11	7.27	+0.16	9.77	8.34	-1.43	7.17	7.86	+0.69
AROTC-248	18.69	17.2	-1.49	7.74	7.4	-0.34	18.71	7.42	-11.29
AROTC-249	8.27	8.24	-0.03	14.24	14.05	-0.19	4.15	8.72	+4.57
AROTC-251	8.19	9.09	+0.90	10.06	10.39	+0.33	8.16	8.08	-0.08
AROTC-252	7.7	8.81	+1.11	12.87	10.43	-2.44	14.42	9.76	-4.66
AROTC-253	1.6	17.95	+16.35	7.99	9.95	+1.96	5.19	5.52	+0.33
AROTC-257	12.32	10.27	-2.05	3.43	3.81	+0.38	3.05	1.26	-1.79
AROTC-261	9.12	13.49	+4.37	7.52	7.6	+0.08	11.71	7.41	-4.3
AROTC-263	6.76	7.03	+0.27	4.65	5.47	+0.82	1.69	0.95	-0.74
AROTC-264	10.47	9.68	-0.79	5.67	7.43	+1.76	8.57	8.37	-0.2
AROTC-266	11.95	13.24	+1.29	5.98	6.78	+0.8	5.91	5.34	-0.57

Individual subject's ankle stiffness (*Kankle*), knee stiffness (*Kknee*), and hip stiffness (*Khip*) during rested and fatigued state, and the change between the two conditions.

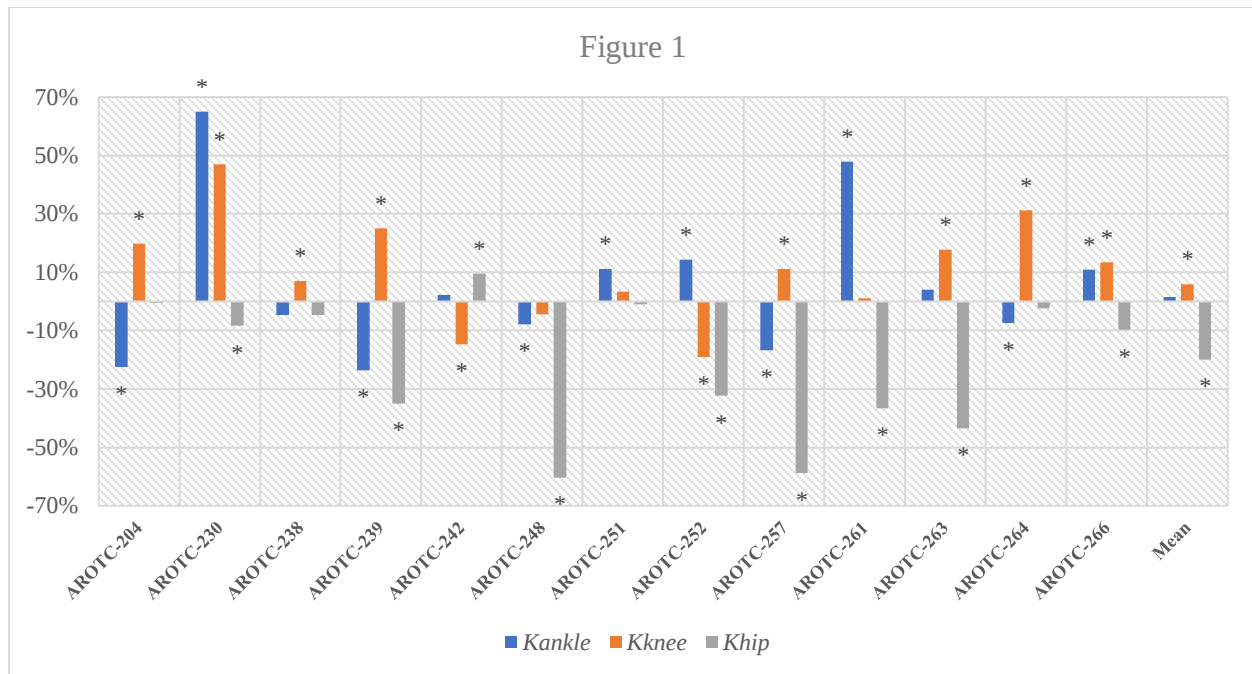


Figure 1. Percent changes for ankle stiffness (*Kankle*), knee stiffness (*Kknee*), and hip stiffness (*Khip*) of individual subjects and group means. Outlier subjects AROTC-240 (+1018% *Khip*), AROTC-249 (+110% *Khip*), and AROTC-253 (+1022% *Kankle*) were excluded from the figure.

* Clinically relevant (> ±5%)

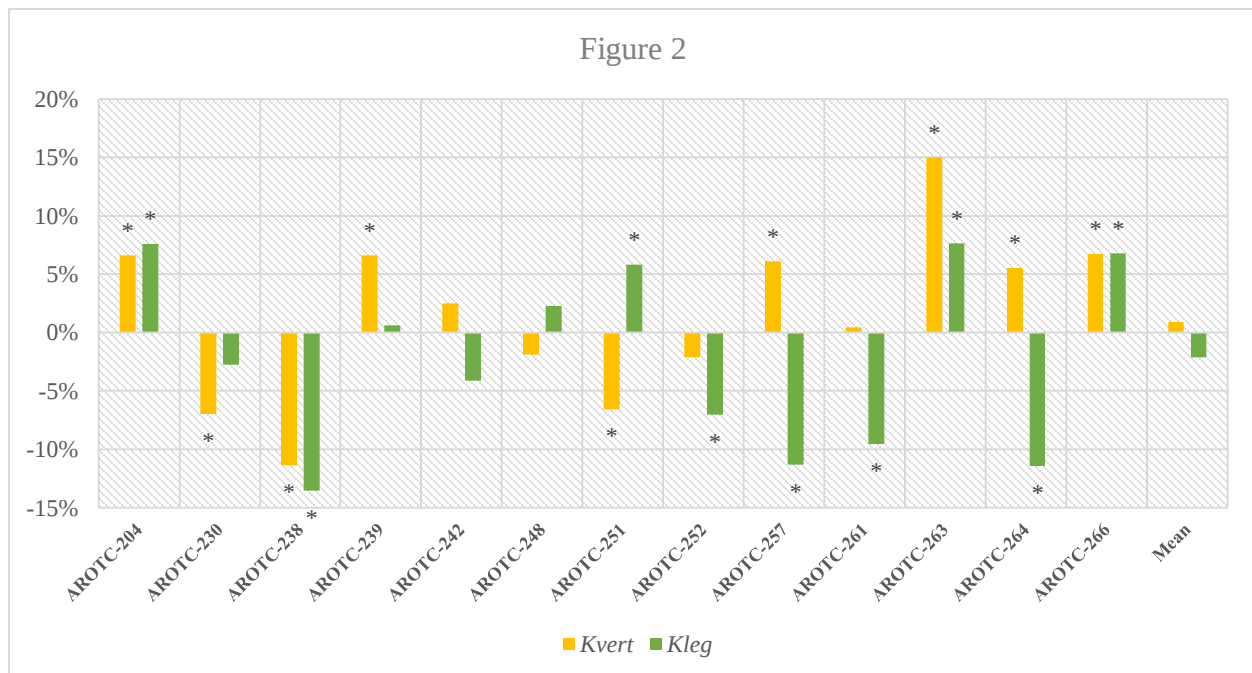


Figure 2. Percent changes for vertical stiffness (*Kvert*) and leg stiffness (*Kleg*) of individual subjects and group means. Outlier subjects were excluded from the figure.

* Clinically relevant (> ±5%)

Discussion

This study investigated the effects of a graded exercise test and exhaustive run at 80% VO₂max treadmill speed on stiffness during running. No changes to stiffness were observed from rested to fatigued state. The hypothesis that K_{vert} would not change is supported by the findings of the current study, three studies that involved steady state runs at VO₂max velocity to volitional exhaustion,^{13,20,21} and one study that involved a 60 minute time trial run at a self-selected pace.¹⁰ Lack of K_{vert} change in the current study was due to no change to F_{max} and Δy . However, the hypothesis that K_{leg} would decrease is not supported by the findings of this study and disagrees with the previously mentioned studies that found no change to K_{vert} .^{10,13,20,21} Although a significant decrease in L_0 and L_c was observed, there was no effect to the ΔL component of K_{leg} since L_0 and L_c decreased proportionally to each other. The decreased L_0 may have been related to greater, although insignificant, lower extremity joint flexion angles at initial contact. Additionally, no changes were observed to other components (θ_L , u , t_c) used to determine ΔL .

The second hypothesis stating that K_{ankle} and K_{knee} would decrease while K_{hip} would not change is partially supported by the findings of the current study. No statistically significant changes were observed to K_{ankle} or K_{knee} group means. Lack of K_{hip} group mean change supported the second hypothesis. Despite an increased $\Delta\theta_{hip}$, K_{hip} likely was not affected due to a slightly reduced ΔM_{hip} . While no significant group changes occurred to K_{joint} , clinically relevant trends existed among some subjects. Half of the subjects showed increased K_{knee} and decreased K_{hip} trends, while some subjects showed increased K_{ankle} and others decreased K_{ankle} trends.

The clinically relevant K_{joint} trends from rested to fatigued state could explain a mechanism for how some subjects attempted to maintain K_{leg} within an optimal zone. Weir et al.²² attributed a lack of K_{leg} change under fatigue to an inverse change between the ankle and knee, where K_{ankle} decreased and K_{knee} increased. In agreement to Weir et al.,²² subjects of the current study trended toward an increased K_{knee} during the fatigued state despite increased $\theta_{knee IC}$ and $\theta_{knee AP Zero}$. However, since $\theta_{knee IC}$ and $\theta_{knee AP Zero}$ increased somewhat proportionally to each other, $\Delta\theta_{knee}$ did not change. It is possible the increased knee flexion angles experienced during the absorption phase of stance had increased the ground reaction force moment arm at the knee, which could have been the cause for the significantly increased $M_{knee AP Zero}$ and ΔM_{knee} . This theory is supported by Sanno et al.³⁵ who found the moment arm at the knee and hip joints increased during a 10km treadmill run. Sanno et al.³⁵ hypothesized the ankle plantarflexor musculature of novice runners fatigued at a faster rate than the knee and hip extensor musculature, which resulted in decreased joint work contributions from the ankle and increased joint work contributions from the more proximal joints.³⁵ Similar joint work changes due to plantarflexor fatigue were observed by Willer et al.³⁶ who also saw increased lower limb flexion angles at initial contact. Increased lower limb flexion angles at initial contact were attributed to later timing of peak hip flexion during swing and shorter flight time.³⁶ The shorter flight time may have been due to reduced ankle power during propulsion from the previous step.³⁶ Later timing of peak hip flexion in combination of a shorter flight time would reduce a runner's ability to extend the lower limb joints and achieve an appropriate L_0 . Subjects of the current study were likely subjected to similar plantarflexor fatigue and reduced ankle power as subjects from Sanno et al.³⁵ and Willer et al.³⁶ We theorize the increased $\theta_{knee IC}$, and the significantly decreased L_0 , could be due to reduced ankle propulsive power from the previous step and shorter flight time.

To maintain K_{leg} in an optimal zone, subjects likely decreased K_{hip} to accommodate the increased K_{knee} . This was achieved by greater $\Delta\theta_{hip}$ for most subjects and decreased ΔM_{hip} for half of the subjects. By decreasing K_{hip} with greater $\Delta\theta_{hip}$ under a fatigued state, subjects may have been shifting the work proximally from the fatigued knee extensors to the stronger and less fatigued hip extensors.

Ankle stiffness may have served as a secondary modulator of K_{leg} since changes were not uniform across subjects. This could be an indication that the magnitude of ankle plantarflexor muscle fatigue varied among the subjects, potentially due to individual differences in fitness or preferred foot strike pattern. The greater dorsiflexion $\theta_{ankle\ IC}$ and plantarflexion $M_{ankle\ IC}$ indicates that absorption at initial contact was shifted toward the ankle dorsiflexor muscles and away from the fatigued ankle plantarflexor muscles.

Maintaining an appropriate stiffness contribution balance between the lower extremity K_{joints} could be an important factor to maintaining a K_{leg} associated with greater running economy. Lower K_{leg} has been associated with reduced running economy since a more compliant leg spring is inefficient at converting potential elastic energy stored during the absorption phase into kinetic energy during the propulsion phase.^{23,24} Tam et al.³⁷ found K_{ankle} was positively correlated and K_{knee} was negatively correlated with oxygen cost during running. The findings of Tam et al.³⁷ support the fact that a lower K_{knee} and low COM style of running, termed “Groucho running,” is associated with reduced running economy due to a higher metabolic demand from increased knee flexion angle and knee extensor muscle work.^{24,35} If joint excursion is the main modulator of K_{joint} , the benefits of lower K_{ankle} for greater running economy could be related to the higher ankle joint work contributions to total lower limb work during economical running or triceps surae morphological characteristics of trained runners.^{38–40}

No relationship between *Khip* and running economy has been established to our knowledge. The findings that *Kknee* increased for half of the subjects, despite trends of increased $\theta_{knee\ IC}$ and $\theta_{knee\ AP\ Zero}$, displays an attempt to utilize an economically advantageous method to ensure that *Kleg* did not decrease under fatigue as observed in previous studies.^{9,10,13,16,20,21} A clear inverse relationship between *Kknee* and *Kankle* was not observed in the current study as was found by Weir et al.²² The current study's findings show that an inverse relationship existed between *Kknee* and *Khip*, however. Had subjects decreased *Kleg* under fatigue because of decreased *Kknee* and *Kankle*, the energy cost of running could have been higher and therefore running economy would be lower.

It should be mentioned that while group mean changes for *Kvert* and *Kleg* were insignificant, there are varied changes among the subjects. We hypothesize that subjects attempted to maintain these stiffness characteristics within an optimal zone to avoid an increase in oxygen cost. Varied changes among the current study's subjects could be attributed to the potential for a wide range of training experience and activity levels despite the sample having been pulled from the same AROTC unit. Similar to the findings of the current study, three studies reported varying *Kvert* and *Kleg* changes among their subjects^{2,9,28} which could be an indication that runners fatigue at different rates from each other or use different neuromuscular strategies to modulate fatigue and energy demand during running. Additionally, the varying range of effects of fatigue on *Kvert* and *Kleg* in research somewhat supports the variation observed in the current study and three studies that found similar variation. Along the polar ends of the running task spectrum, from sprints to ultraendurance runs, there is a common pattern to effects of neuromuscular fatigue on *Kvert* and *Kleg*. From sprint running-induced fatigue, *Kvert* tended to decrease and *Kleg* did not change.^{12,14,19} During ultraendurance runs, *Kvert* tended to

increase and *Kleg* increased or remained the same.^{8,17,18} Varying changes to *Kvert* and *Kleg* appeared in exhaustive runs that fit in the middle of the spectrum between sprints and ultraendurance runs. It appears that *Kvert* either decreased or remained the same,^{11,13,15,16,20,21,29} and *Kleg* tended to decrease^{13,16,20,21} during shorter and higher intensity runs at fixed velocities, such as at VO2max velocity. Among slightly longer and lower intensity steady state runs, *Kvert* remained the same or increased^{2,3,10,28} and *Kleg* remained the same or decreased.^{9,10,22,28} Clearly, the type of fatigue protocol and running task are large factors that could affect stiffness modulation. Varying results among different studies could also be due to the subjects used and different methods for stiffness calculations. There does not seem to be a clear trend for fatigue effects on *Kvert* or *Kleg* during running between runners of differing experience and training status. A study that makes a direct comparison between groups of different experience levels may be necessary to establish that relationship.

Several limitations exist from the current study. First, the sample was limited to sixteen subjects from a large pool of AROTC cadets at the university level. Although the subjects endure similar training as part of being a member of AROTC, prior training experience and current activity levels were not completely controlled as was done for other studies that investigated fatigue and stiffness. Because of this, we cannot use these findings to represent a large population of trained runners. Had the sample been larger, the findings may have been representative of a physically active general population that does not solely train running. Second, the fatigue protocol was conducted on a treadmill and included increasing grade every three minutes, which may not be representative of regular running training if subjects spend most of their time running over flat ground. Lastly, the only spatiotemporal parameter collected was

ground contact time (t_c). We could have described fatigue's effects on the stiffness variables in more detail had flight time, step frequency, and step length been collected.

Conclusion

To our knowledge, this is the first study to investigate fatigue effects on K_{vert} , K_{leg} , and K_{joint} of the ankle, knee, and hip. Fatigue did not induce changes to K_{vert} , K_{leg} , or K_{joint} group means. However, varying K_{vert} and K_{leg} changes existed among subjects, and clinically relevant trends existed for K_{joint} changes among subjects. To maintain K_{leg} , some subjects increased K_{knee} and decreased K_{hip} while modulating K_{ankle} on a secondary level. The observed K_{joint} trends could be a strategy to avoid stiffness characteristics associated with increased oxygen cost during running while fatigued.

Review of Literature

Neuromuscular Fatigue

Neuromuscular fatigue occurs due to repetitive stretch-shortening of musculotendinous units.⁵ Mechanisms behind neuromuscular fatigue include metabolic substrate depletion or structural damage to muscle tissue.⁵ Structural damage to tissue can reduce passive musculotendinous unit compliance and stretch-reflex characteristics,^{5,6,41} resulting in a loss in maximum force production during voluntary muscle contraction or failure to maintain a given exercise intensity.^{1,5} Researchers investigating neuromuscular fatigue utilize fatigue protocols such as isolated maximum voluntary isometric and isokinetic muscle contractions,^{42,43} bodyweight or loaded dynamic exercises,^{44–46} shuttle runs or sprints,^{12,14,19,40,46} or exhaustive runs.^{1–3,7–11,13,15–18,20–22,28,29,35–37,46–48} Due to differences in onset of fatigue and fatigue severity between individuals, it has been suggested that maximum or near maximum fatigue protocols can elicit true neuromuscular performance deteriorations, whereas submaximal fatigue protocols may only elicit facilitation of neuromuscular performance.⁵

Reductions in neuromuscular performance due to fatigue are often observed in maximum voluntary isometric contractions (MVIC),^{4,6,40} or dynamic exercises such as jumping,¹ drop landings,^{44–46} and running.^{1,12,14,19,35,48} Kwon and Williams⁴⁹ observed localized muscle fatigue of the ankle plantarflexors and dorsiflexors following a Bruce protocol. Fatigue was indicated by a 20% decrease in plantarflexion moment and 12% decrease in dorsiflexion moment during MVIC.⁴⁹ Running induced ankle plantarflexor fatigue is expected due to the relatively higher demand during running compared to knee and hip extensor muscles.³⁵ The smaller muscle groups surrounding the ankle joint that produce the majority of power and work for propulsion

can fatigue at a faster rate than the larger muscle groups of the knee and hip due to differences in strength and work capacity.³⁵

Potential differences in fatigue rate of different lower limb muscle groups have been indicated by changes in positive joint work during high intensity prolonged runs.^{35,36,48} Sanno et al.³⁵ investigated relative joint work contributions to total lower extremity work in recreational and competitive runners during a 10-km treadmill run. Subjects ran at a pace of 105% of their best 10 km time. Recreational runners experienced a proximal shift in positive joint work contributions due to -14% decreased positive ankle joint work, +14% increased positive knee joint work, and +22% increased hip joint work. Competitive runners, however, did not experience the same proximal shift in joint work contributions. The competitive runners' ankle joint work only decreased -6%, knee joint work increased +13%, and hip joint work only increased +1%. The competitive runners in this study likely had a higher work capacity for their ankle plantarflexors and an increased plantarflexion velocity strategy may have reduced work contribution changes that could have occurred due to fatigue.³⁵ In a similarly designed study, Sanno et al.⁴⁸ investigated the differences between a racing flat shoe and a cushioned shoe's influences on joint work changes using the same fatigue protocol as Sanno et al.³⁵ In agreement to Sanno et al.'s³⁵ recreational runner subjects, Sanno et al.⁴⁸ found a similar proximal shift in relative joint work contributions among their competitive and recreational runners for both shoe conditions. In contrast, Melaro et al.⁴⁷ did not observe a proximal shift in relative joint work contributions to total lower limb work despite a moderate, yet insignificant, reduction in ankle joint work. Lack of changes were attributed to a lower intensity running protocol with the use of subjects' typical running pace used in training.⁴⁷ The use of a lower running intensity protocol could have demonstrated that neuromuscular performance, particularly propulsive performance,

is able to be maintained under conditions similar to a runner's normal training, and that higher intensity fatigue protocols elicit performance reductions.⁵

Running induced neuromuscular fatigue can also affect absorption strategies. It has been observed that under fatigue, runners land with a more flexed lower limb due greater knee flexion at initial contact.^{29,35} Möhler et al.¹⁶ observed greater sagittal plane ranges of motion of the ankle and knee while Luo et al.¹⁵ and Möhler et al.²⁹ observed greater sagittal plane ranges of motion of the knee and hip. Although ranges of motions change during the absorption phase of running, relative joint work contributions to total negative work of the lower limb do not appear to change according to the studies by Sanno et al.,³⁵ Melaro et al.,⁴⁷ and Sanno et al.⁴⁸ Because of this, it may be better to look at spring-mass characteristics to understand how absorption strategies are effected by neuromuscular fatigue.

Spring-Mass Model

The human running motion has been described using a basic spring-mass model.²³⁻²⁵ The spring-mass model consists of a single point mass supported by a massless linear spring. The spring-mass system can undergo vertical oscillation involving an absorption phase and propulsion phase. Downward compression of the system represents absorption and storage of potential energy, and upward recoil represents potential energy conversion into kinetic energy for propulsion. In running, the lower limb acts as the spring, or “leg spring”, supporting the center of mass (COM) through the stance phase. The first half of the stance phase is the absorption phase where the leg spring compresses to absorb external forces and store potential elastic energy in viscoelastic components such as muscles, tendons, and ligaments. At midstance, it is assumed that the leg spring reaches full compression, and the COM reaches its lowest vertical position. The second half of the stance phase is the propulsion phase where the leg

spring's potential energy is converted into kinetic energy to propel the COM. During hopping or running tasks, spring-mass characteristics such as ground contact time, flight time, frequency, and spring stiffness are related to each other and can be analyzed to understand the neuromuscular modulation of the spring-mass system.

Stiffness

Spring stiffness originates from physics' Hooke's Law which is defined by the equation $F=kx$, where F is the deforming force, k is the spring constant (stiffness), and x is the distance the spring deforms.²³ Therefore, k is the ratio between F and x from the rearranged equation $k=F/x$. Vertical stiffness (K_{vert}) uses Hooke's law directly with the equation $K_{vert}= F_{max}/\Delta y$, where F_{max} is the maximum vertical ground reaction force of the stance phase and Δy is the COM vertical displacement from ground contact to the lowest position achieved during stance.^{23,24} During running, K_{vert} describes the leg spring's compliance and COM trajectory during the stance phase. A runner with a greater K_{vert} will be characterized by a higher F_{max} or lower Δy , and a runner with a lower K_{vert} will be characterized by a lower F_{max} or greater Δy .

Vertical stiffness is limited to describing vertical oscillation due to its one-dimensional calculation. Running, however, involves a horizontal component. The leg spring contacts the ground at an angle, and the COM moves horizontally while simultaneously undergoing vertical oscillation. As a result, the COM has a sinusoidal motion pattern. Leg stiffness (K_{leg}) is the proposed method of accounting for the lower limb's contact angle with the ground.²³ Leg stiffness is calculated using the equation $K_{leg} = F_{max}/\Delta L$, where F_{max} is the maximum vertical ground reaction force and ΔL is the change in vertical leg length.³³ Change in vertical leg length is calculated using the equation $\Delta L = \Delta y + L_0(1 - \cos\theta)$, where Δy is COM displacement, L_0 is leg length at initial contact.³³ θ is the half angle of the arc swept by the leg and is calculated with the

equation $\theta = \sin^{-1}(ut_c/2L_0)$, where u is the forward velocity of center of mass and t_c is ground contact time.³³

Leg stiffness is modulated by a torsional spring system of the lower limb.^{23,24} The sum of joint stiffnesses (K_{joint}) of the ankle, knee, and hip joints make up the torsional spring system. Joint stiffness describes spring absorption characteristics from initial contact to either midstance,^{15,22} the time when maximum joint flexion occurs during stance,⁵⁰ or when the anterior-posterior ground reaction force vector is zero.^{34,51} Joint stiffness is calculated with the equation $K_{joint} = \Delta M / \Delta \theta$, where ΔM is the change in external joint moment and $\Delta \theta$ is the angular joint displacement during the absorption phase.^{15,22,34,50,51}

Differences in K_{joint} among the ankle, knee, and hip could indicate a difference in relative stiffness contributions to K_{vert} or K_{leg} . Farley & Morgenroth⁵² concluded that K_{ankle} had the greatest influence on K_{leg} compared to K_{knee} and K_{hip} during submaximal hopping. Hobara et al.⁵³ observed that K_{knee} had the greatest influence on K_{leg} during maximal hopping. Across varying running velocities, Arampatzis et al.⁵⁴ determined that K_{knee} had the greatest influence on K_{leg} . As running velocity increased, K_{vert} , K_{leg} , and K_{knee} increased whereas K_{ankle} did not change. Although Arampatzis et al.⁵⁴ did not explicitly mention that K_{knee} had a greater influence on K_{vert} than K_{ankle} , this likely was the case due to a similar increase for K_{vert} and K_{leg} as velocity increased.

In general, K_{vert} increases with the demand of the activity, such as increasing running velocity, to allow for an efficient recoil of propulsion after absorption of ground reaction forces.^{23,24,54} While K_{vert} increases, ground contact time decreases, flight time increases, and step frequency increases.²⁴ These characteristics combined during steady state running are also associated with greater running economy due to a better ability to absorb ground reaction forces,

store elastic potential energy in musculotendinous units, and return the elastic potential energy as kinetic energy.²⁴ Those with a more compliant limb have been observed to have a reduced running economy, likely due to a greater metabolic demand associated with an inefficient transition from absorption to propulsion.^{24,37}

Stiffness and Fatigue

There appears to be differences in the effects of neuromuscular fatigue on stiffness changes, or lack thereof. Ideally, a runner would attempt to maintain spring-mass characteristics associated with greater running economy during prolonged runs. During prolonged runs however, neuromuscular fatigue can limit the capacity for muscles to work efficiently and maintain performance. Given that some studies have observed increases in lower limb joint ranges of motion due to fatigue,^{15,16,29,35} reduced *Kvert*, *Kleg*, and *Kjoint* are plausible.

Dutto & Smith⁹ investigated the effects of fatigue on well-trained runners' *Kvert* and *Kleg* induced by an exhaustive treadmill run at a speed associated with 80% of subjects' VO₂max. Subjects ran to volitional exhaustion at 4.03 ± 0.36 m/s for a duration of 57 ± 19 min. Twelve subjects decreased *Kvert* up to -8.7% and two subjects increased *Kvert* up to +6%. Changes in *Kvert* were inversely related to changes in Δy and proportional with changes in stride frequency. Most subjects experienced changes in *Kleg* up to 13% due to ΔL , however the authors did not expand on observations between subjects that decreased, increased, or maintained *Kleg*.⁹ Luo et al.¹⁵ observed *Kvert* insignificantly decreased from 67.41 ± 10.54 N/m/kg to 61.32 ± 6.67 N/m/kg in recreational runners during a 3.33m/s exhaustive treadmill run that lasted an average of 28.5 minutes. In agreement with Dutto & Smith,⁹ Luo et al.¹⁵ attributed *Kvert* decreases to increased Δy . During an exhaustive treadmill run at VO₂max velocity, Hayes & Caplan¹³ observed six sub-elite middle distance runners insignificantly decreased *Kvert* from $72410 \pm$

18740 N/m to 66486 ± 15359 N/m due to a small increase in Δy , and significantly decreased K_{leg} from 6403 ± 1486 N/m to 5306 ± 925 N/m due to a significant increase in ΔL and decrease in ground contact time. Among these studies that observed decreased K_{vert} and K_{leg} , changes in Δy and ΔL appear to be the primary factor to stiffness changes compared to changes in F_{max} .^{9,13,15}

It is possible to change stiffness properties by altering the F_{max} during running. In a study by Girard et al.,¹¹ 12 regional and national level triathletes decreased K_{vert} by -6% during a 5-km time trial run (~ 17 min 31 s, 17.3 ± 0.3 km/h) on an indoor track due to decreased F_{max} . Leg stiffness, however, was maintained due to proportionately increased ΔL and decreased F_{max} .¹¹ Möhler et al.¹⁶ investigated kinematics, K_{vert} , and K_{leg} in 13 trained runners before and after a treadmill run protocol designed to exhaust subjects at around 10 minutes. Subjects ran to volitional exhaustion at velocities associated with 110% of their velocity at 4 mmol/l lactate, which averaged 19.27 ± 0.72 km/h. Vertical stiffness decreased from 20.55 ± 3.98 kN/m to 18.01 ± 4.56 kN/m and K_{leg} decreased from 12.40 ± 2.62 kN/m to 10.56 ± 2.90 kN/m. Möhler et al.¹⁶ attributed stiffness decreases to increased ground contact time. Leg stiffness was likely influenced by increased sagittal plane range of motion at the ankle, knee, and hip joints as well. Center of mass displacement decreased despite the increased ranges of motion of the lower limb.¹⁶ If Δy decreased, there must have been changes in F_{max} to cause the decreased K_{vert} . However, Möhler et al.¹⁶ calculated K_{vert} kinematically without collecting F_{max} , so this assumption is not certain.

Observed K_{vert} decreases during exhaustive runs due to increased COM^{9,13,15} are in agreement with studies investigating K_{vert} changes during sprint running.^{12,14,19} Neuromuscular fatigue from 12 repeated 40-m sprints,¹² four repeated 100-m sprints,¹⁹ and one 400m sprint¹⁴

caused K_{vert} to decrease gradually due to increased Δy instead of decreased F_{max} . Additionally, K_{vert} decreases were associated with decreased running velocity, decreased step and stride frequency, and increased ground contact time.^{12,14,19} Inability to control Δy with decreasing K_{vert} during sprints and moderate to high-intensity prolonged runs demonstrates how neuromuscular fatigue can negatively affect running performance. Despite steady decreases in K_{vert} , K_{leg} remained constant or changed very little due to no significant changes to F_{max} or ΔL .^{10,15,30}

Opposite to sprinting and previously mentioned prolonged runs to exhaustion, K_{vert} and K_{leg} increased during long distance runs that require extremely efficient running mechanics to endure the repetitive stretch-shortening cycles.^{8,17,18} Morin et al.¹⁷ investigated the effects of a 24-hour treadmill run protocol on K_{vert} and K_{leg} of 10 experienced ultraendurance runners. Spring-mass parameters were collected for 60 seconds just before, every two hours, and after the 24-hour run on a separate treadmill dynamometer that measured ground reaction forces. Other neuromechanical and physiological tests were performed during the run as part of a larger project. Subjects covered an average distance of 153 ± 15 km, and the actual time spent walking and running was 18 hours. Vertical stiffness significantly increased from 29.4 ± 5.54 kN/m to 32.3 ± 4.34 kN/m and K_{leg} significantly increased from 15.5 ± 3.55 kN/m to 16.8 ± 2.53 kN/m. Although F_{max} decreased, stiffness increases were due to larger decreases to Δy and ΔL . Decreased vertical oscillation was associated with increased step frequency and decreased ground contact time. Flight time, however, did not change.¹⁷

In another study, Morin et al.¹⁸ investigated fatigue-induced changes to spring-mass characteristics in 18 experienced ultramarathon runners who ran a 166-km mountain ultramarathon race with 9500-m of positive and negative elevation change. Subjects ran the race in 37.9 ± 6.2 hours. Before and after the race, spring-mass characteristics were collected on a 7.32-

m runway with subjects running overground at 12 km/h. Vertical stiffness significantly increased from 25.1 ± 2.32 kN/m to 26.6 ± 3.32 kN/m due to the same conditions as observed by Morin et al.,¹⁷ whereas *Kleg* insignificantly decreased from 9.87 ± 1.45 kN/m to 9.44 ± 1.10 kN/m due to decreased *Fmax* and no change to ΔL .¹⁸ In contrast to the 24-hour treadmill run by Morin et al.,¹⁷ ground contact time did not change and flight time decreased.¹⁸ This resulted in a similar increase to step frequency, but through a different strategy.¹⁸

Degache et al.⁸ investigated the effects of a five-hour hilly run on *Kvert* and *Kleg* of eight regional level male long distance runners. The subjects ran on a 1.7 km hillside circuit with 75-m of positive and negative elevation change and covered a distance of 37.5 ± 5.5 km and 1730 ± 230 m of elevation change during the five hours. Before and immediately after the run, spring-mass characteristics were collected on a treadmill dynamometer for one minute at 10, 12, and 14 km/h with one minute rest between each velocity. Although a much shorter distance covered than the 24-hour treadmill run from Morin et al.,¹⁷ subjects similarly increased *Kvert* due to decreased Δy , increased *Kleg* due to decreased ΔL , and increased step frequency due to decreased ground contact time and no change to flight time.⁸

An ample number of studies have observed subjects maintaining *Kvert* during exhaustive runs. Möhler et al.²⁹ investigated kinematics, *Kvert*, and *Kleg* of 14 novice runners during a 13 km/h treadmill run to exhaustion. Subjects lasted an average of 6.18 minutes. While the sagittal knee and hip range of motion increased significantly, sagittal ankle and COM range of motion increased insignificantly. The novice runners were able to maintain stride frequency, *Kvert*, and *Kleg* throughout the exhaustive run,²⁹ which disagreed with a previous study on expert runners.¹⁶ This could have been due to novice runners from Möhler et al.²⁹ not having the experience to

endure the fatigue protocol to reach similar durations and fatigue severity like the trained runners from Möhler et al.¹⁶

Two studies that investigated the effects of fatigue on *Kvert* from running at a velocity associated with VO_2max ²¹ and 95% of VO_2max velocity²⁰ found no change to *Kvert* and decreased *Kleg*. Rabita et al.²⁰ had nine national level triathletes run around a 200-m indoor track at 95% of VO_2max velocity to exhaustion (353 ± 69 s, 1780 ± 317 m). Rabita et al.²¹ had 12 trained runners run around a 340-m indoor track at VO_2max velocity to exhaustion (10.7 ± 2.9 min, 3258 ± 819 m). Both studies observed increased ground contact times, no change in *Kvert* due to proportional decreases in both Δy and *Fmax*, and decreased *Kleg* 11.7 ± 1.9 kN/m to 10.4 ± 1.0 kN/m²⁰ and 13.9 ± 3.3 kN/m to 12.6 ± 2.9 kN/m²¹ due to the decreased *Fmax*. Rabita et al.²¹ observed increased leg compression whereas Rabita et al.²⁰ did not. Interestingly, Rabita et al.²⁰ observed increased step frequency throughout the run despite the increased ground contact time, and attributed those findings to the fact that flight time decreased to a larger degree than ground contact time increased. Rabita et al.²¹ did not observe changes in step frequency, and this was attributed to a proportionately decreased flight time and increased ground contact time.

In agreement to the two studies just described,^{20,21} Garcia-Pinillos et al.¹⁰ observed no change to *Kvert* and decreased *Kleg*, but during a much longer running protocol. In this study, trained endurance runners ran a 60-minute time trial on a treadmill at a self-selected pace (15.1 ± 0.6 km/h) and ended with an RPE of 19.3 ± 0.9 according to Borg's 6-20 scale. Kinematics were captured before and after the time trial on the treadmill at 12 km/h for a duration of three minutes. Stiffnesses were kinematically calculated using the method from Morin et al.⁵⁵ Leg stiffness decreased from 8.62 ± 1.50 kN/m to 7.86 ± 1.33 kN/m. Ground contact time and step frequency increased, and flight time and step length decreased.¹⁰ The authors did not report

observations to Δy or ΔL , so inferences to why K_{vert} was maintained and why K_{leg} decreased cannot be made.

A potential explanation for how K_{vert} was maintained with a decreased K_{leg} and increased ground contact time among Rabita et al.,²¹ Rabita et al.,²⁰ and Garcia-Pinillos et al.¹⁰ is that subjects ran with the “Groucho style”, which involves a low COM via a compliant lower limb.^{24,35} To further explain, the subjects could have had a greater angle of the lower limb during initial contact due to reaching further out in front during swing. This would result in a lower COM during initial contact. During the first half of stance, the leg could have compressed significantly more as the COM traveled forward. While considered inefficient for prolonged running due to a higher energy demand,³⁵ the combination of greater lower limb contact angle and increased leg compression over the first half of stance could have resulted in the decreased Δy and F_{max} .^{10,20,21}

Maintaining or increasing running efficiency over prolonged runs also includes symmetry between limbs. Putnam² and Radzak et al.³ investigated K_{vert} and knee K_{joint} symmetry between limbs. Both studies involved cadets from a University Reserve Officer’s Training Corp that ran a Modified Åstrand GXT to determine VO₂max, followed by a prolonged run to volitional exhaustion at a treadmill velocity intended to elicit approximately 80% of VO₂max. Biomechanics data were collected prior to and immediately after the fatigue protocol on an 18-m runway at 4.0 m/s $\pm$ 10%. Putnam² categorized limbs as more stiff or less stiff from rested state K_{vert} , regardless of being right or left limbs. The limbs that were less stiff under a rested state increased K_{vert} (51.33 $\pm$ 6.60 kN/m to 55.53 $\pm$ 8.49 kN/m) and the stiffer limbs under a rested state did not change K_{vert} (58.95 $\pm$ 9.95 kN/m to 58.61 $\pm$ 10.13 kN/m). This resulted in better K_{vert} symmetry between limbs under the fatigued state. Vertical ground reaction force and

ground contact time did not change for either limb. Increased K_{vert} of the less stiff could have been due to a decreased Δy , however COM kinematics was not reported. Both limbs decreased K_{knee} from rested to fatigued state, however the change was insignificant. Small K_{knee} decreases could be explained by a significant decrease of the less stiff limb's knee flexion moment and near significant decrease of the stiffer limb's knee flexion moment.²

Radzak et al.³ investigated symmetry between right and left limbs and found that in a rested state, the right limb's K_{vert} was less stiff than the left limb. Similar to Putnam,² the stiffer limb (left limb) did not change K_{vert} (40.66 ± 9.21 kN/m to 40.48 ± 7.73 kN/m), the less stiff limb (right limb) increased K_{vert} (37.33 ± 7.21 kN/m to 39.52 ± 8.51 kN/m), and K_{vert} between limbs became more symmetrical.³ Increased K_{vert} of the right limb could be attributed to increased F_{max} of the right limb. For the left limb, which did not change K_{vert} , F_{max} did not change. In contrast to K_{vert} symmetry, K_{knee} became more asymmetrical due to fatigue. Interestingly, the less stiff right limb which increased K_{vert} had no change to K_{knee} (4.93 ± 4.00 Nm/deg to 4.39 ± 2.96 Nm/deg), and the stiffer left limb which did not change K_{vert} had decreased K_{knee} (4.09 ± 2.94 Nm/deg to 2.96 ± 1.76 Nm/deg). Increased K_{ankle} and K_{hip} are a potential joint level mechanisms to explain why right limb K_{vert} did not change while K_{knee} decreased, and why left limb K_{vert} increased with no change to knee K_{joint} .³ However, K_{ankle} and K_{hip} was not collected, so it is unknown if these changes occurred.

Joint level stiffness contributions to K_{leg} from rested to fatigued state have been investigated by Weir et al.²² Recreational runners completed runs on two separate days at a self-selected pace (3.33 ± 0.4 m/s). One data collection day involved a 21-minute run in a neutral shoe followed by another 21 min run in a different neutral shoe. The other data collection day involved a 21-minute run in a neutral shoe followed by another 21-minute run in a stability shoe.

Leg stiffness did not change during either neutral-neutral shoe or neutral-stability shoe runs. In contrast to Radzak et al.,³ K_{knee} increased from 6.54 ± 1.25 Nm/deg to 6.86 ± 1.21 Nm/deg for the neutral-neutral shoe run and increased from 6.67 ± 1.39 Nm/deg to 6.92 ± 1.35 Nm/deg for the neutral-stability shoe run. Inversely, K_{ankle} decreased from 12.13 ± 2.32 Nm/deg to 10.76 ± 2.98 Nm/deg for the neutral-neutral shoe run and from 11.27 ± 2.23 Nm/deg to 10.88 Nm/deg for the neutral-stability shoe run. Lack of K_{leg} changes were attributed to the inverse K_{ankle} and K_{knee} changes.²² It is unknown if the hip influenced K_{leg} since K_{hip} was not collected.

Joint stiffnesses of the ankle, knee, and hip were investigated in a study mentioned earlier by Luo et al.¹⁵ where K_{vert} insignificantly decreased under fatigue due to increased Δy . In agreement with Weir et al.,²² K_{ankle} significantly decreased (0.21 ± 0.03 Nm/deg/kg to 0.20 ± 0.04 Nm/deg/kg) and K_{knee} significantly increased (0.04 ± 0.03 Nm/deg/kg to 0.06 ± 0.03 Nm/deg/kg). A trending increase to K_{hip} (0.55 ± 0.35 Nm/deg/kg to 0.77 ± 0.45 Nm/deg/kg) was observed, but it was not significant. Ankle and knee joint excursion from initial contact to midstance did not experience significant changes while hip joint excursion increased significantly. Changes in joint moment for ankle, knee, and hip joint were not reported.¹⁵

Understanding K_{joint} contributions to K_{leg} and K_{vert} can improve understanding of neuromuscular fatigue effects on spring-mass characteristics. Inverse K_{joint} relationships between the ankle and knee have been one method of describing K_{leg} modulation under fatigue.²² How K_{hip} fits into that relationship is unclear due to limited research investigating fatigue and K_{hip} . The fatigue induced K_{ankle} decrease and K_{knee} increase observed by Weir et al.²² and Luo et al.¹⁵ could be an indication of an altered absorption strategy where more musculotendinous unit absorption takes place at the ankle joint and less at the knee joint. It is well known that a running style with a compliant knee joint, “Groucho Running”, requires a

higher metabolic cost,^{24,35} and decreasing quadriceps muscle absorption through a stiffer knee joint during a prolonged run may be a strategy to reduce the metabolic demand. Through a more compliant ankle joint, increasing absorption under fatigue may be a strategy to increase the stretch of the triceps surae and Achilles tendon musculotendinous unit to store more potential elastic energy for propulsion.

Increasing or maintaining *Kvert* and *Kleg* appears to be favorable during moderate to ultra-long distance running protocols and races. Spring-mass characteristics such as decreased Δy and ΔL , decreased ground contact time, increased flight time, and increased step frequency are related to the increased *Kvert* and *Kleg*.^{8,17,18} These changes in line with increased stiffness were hypothesized to be a “smoother” and “safer” style of running which reduced the eccentric muscle work required for long distance runs.¹⁷ The potentially reduced eccentric muscle work from increased *Kvert* and *Kleg* could be in agreement with increased *Kknee* of Weir et al.²² and Luo et al.¹⁵ as a strategy to decrease the metabolic demand of running.

Differences in fatigue induced effects on stiffness parameters among the studies mentioned are likely attributed to differences in methodology. First, most of the studies vary in running fatigue protocols, ranging from 40-m sprints to 166-km runs.^{1–3,8–22,28,29} Levels of fatigue and types of fatigue induced by the wide range of protocols likely vary. Additionally, the method of data collection among the studies varied. Some studies collected overground running trials before and after a treadmill run or overground race.^{1–3,18} Other studies collected running biomechanics during a portion of the beginning and end of a treadmill run,^{8,10,13,16,22,28,29} or throughout a treadmill or overground run.^{9,11–13,15,17,19–21} Vertical ground reaction force was not collected in all studies, which required the researchers to use kinematic methods to calculate *Kvert* or *Kleg*.^{10,13,14,16,18,19,29} And not all studies collected kinematics, so COM trajectory was

estimated by double integration of the vertical ground reaction force curve to determine K_{vert} or K_{leg} .^{11,12,17,28} Using different methods to calculate stiffness is likely a reason for the range of different values among the studies. To compare stiffness changes due to fatigue among different studies, it is important to understand the context that fatigue was induced and how the results were calculated.

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