AN AGE-BASED KINETIC ANALYSIS THAT UNDERPINS JOINT COORDINATION IN RUNNING PERFORMANCE

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The purpose of this study was to examine the age-based biomechanics that underpins lower body joint coordination during running gait. A longitudinal approach was adopted and experienced male distance runners (initially with a mean age = 53.34 years) were examined over a seven-year period. Three-dimensional joint coordinate and contact ground reaction force data were collected for running trials with a horizontal velocity = 3.83±0.40 $m \cdot s^{-1}$. ANOVA showed significant increases ($p \leq 0.03$) in knee joint moments at amortisation and peak impact and braking forces. Statistical parametric mapping identified decreased hip moments at ~25-40 % of stance. With age there is an increase requirement to attenuate the passive impact forces which places the lower body musculature under further stress. The changes in hip moments, whilst the hip joint is extending, suggests a shift in power production to the knee and ankle which has been previously reported with faster running (Ferris et al., 2012).

KEYWORDS: age, longitudinal study, running gait, kinetics

INTRODUCTION: Examining the underpinning biomechanics of running gait in Master's runners can provide valuable insight and understanding of the effect of age on dynamic performance. During running gait, the initial impact with the ground produces a shock wave up the body where the force attenuation capabilities reduce the impact energy and the amplitude of the loading on the body. Previous studies have suggested that there are contradictory responses in the absorption of the passive contact forces between young and old athletes (Gittoes & Wilson, 2010). Diss et al. (2019) explored the change in tri-joint coordination with age using a contemporary measure of cluster phases analysis. Over a seven year period of ageing in Master athletes synchrony of the sagittal plane angles in all three lower body joints significantly increased in the absorption phase during the stance period of running gait. They concluded that the force attenuation strategies are comprised with age and the increase in tri-joint synchrony was a mechanism to minimise injury during the loading phase of running.
Understanding the changes of the lower body joint moments is merited to gain insight of their response to ageing. Kuitunen et al. (2002) reported decreased joint moments with age which was attributed to a reduction in the capacity to tolerate the applied load during stance. An age-based cross-sectional analysis reported a predominately reduced sagittal plane ankle joint moment during stance in older athletes (Diss et al., 2015) which provided further support for the decline in the force-velocity response of the joint musculature with age.
A decline in self-selected running velocity has been reported with age (Power et al, 2012) which was attributed to a decrease in step length and frequency. However, Diss et al. (2015) reported that a slower gait with age was solely associated with a reduction in step length. Orendurff et al. (2019) examined the sagittal plane lower body moments and the effect of running speed. Increases in speed found an increased knee extensor and plantar-flexor moments whilst the hip flexor moment decreased as the body transitioned from a downward to an upward motion of the centre of mass. They also reported a sequential link of the hip flexor moment to the ankle plantar-flexor moment.
Ageing research examining dynamic movement has been typically adopted a cross-sectional design and has focussed on walking (Lilley et al., 2011). A longitudinal design provides a better understanding of age-based changes in running gait by considering individual changes prospectively. The aim of this study was to examine the age-based kinetics, using a longitudinal approach, that underpin the changes in lower body joint synchrony.

METHODS: Ten male endurance-trained athletes (age = 53.54±2.56 years, mass = 71.05±7.92 kg) volunteered to participate in the study and returned to the study seven years later (age = 60.49±2.56 years, mass = 69.08±8.23 kg). M50 defined the initial data collection and M57 the data collection seven years later. The criterion for inclusion in the study required the athletes to: be injury free, participate in a minimum of five running-based training sessions per week (two of which were at an intensity that exceeded the lactate threshold), have a personal best time for 10 km of less than 40 minutes, finish in the top twenty positions in the regional county championships. All athletes provided written informed consent, and ethical approval for the data collection protocol was gained from the host University's Ethics Board prior to study onset.

Passive markers were placed at precise anatomical landmarks and anthropometric measurements were recorded in accordance with the lower body Plug-in-Gait model (Vicon^{TM}, Oxford). Following a familiarisation period, participants performed multiple running trials (typically 20) at a horizontal velocity = 3.83±0.40 $m \cdot s^{-1}$ whilst making right foot-ground contact with a force plate situated 13 m along the 20 m runway. Three-dimensional coordinate (sample rate: 120 Hz) data of the passive markers were collected using a 12 camera Vicon system (Vicon ^{TM}, Oxford) synchronised with a Kistler force plate (Kistler^{TM}, Switzerland, 9281C; sample rate: 1080 Hz). The protocol and data collection were replicated seven-years later.

The three-dimensional coordinate data time histories were smoothed using Woltring's cross-validated quintic spline with the mean square error noise tolerance level set to 15 mm^2 from which the joint centres of the lower body were determined. Six trials for each participant were used for further analysis. The average horizontal velocity of the centre of mass over one gait cycle determined the running speed and a single-step length was defined by the horizontal displacement of the ankle joint marker between the contralateral foot touch-down events. Subsequentially step frequency was determined. Stance phase ground contact forces (GRF) and joint moments of each running trial were analysed and defined between the instants of initial ground contact (GRFz > 8N) and toe off (GRFz < 8N) with the force plate. The GRF were normalised to body weight (BW) and the moments to BW and leg length. Amortisation was defined as the time when the centre of mass was at a minimum during the stance phase. Individual stance phase waveform profiles of the GRF and joint moments were interpolated to 101 points using a cubic spline (MathCad 13, Adept Scientific). The average of all stance phase measures were calculated for each athlete from the six athlete specific trails for both data collection sessions. The group means (standard deviation) were then determined for the discrete and waveform measures.

The Shapiro-Wilk statistical test for normal distribution revealed that all measures were normally distributed. Statistical parametric mapping (SPM) technique with paired t-test was used to examine the differences in the waveform GRF and moment data for M50 and M57. SPM was designed was designed especially for continuous field analysis (Friston, Ashburner, Kiebel, Nichols & Penny, 2007) and constructs images that lie in the original, biomechanically meaningful sampling space (Pataky, 2010). Open-source one-dimensional package for Matlab (spm1d version M.0>3.1 (2015.08.28) was used in the analysis and the scalar statistic SPM {t} was computed at each point in the time series as described previously by Robinson, Vanrenterghem and Pataky (2015). Analysis of variance (ANOVA) was conducted to determine significant difference between the discrete measures. Percentage difference were calculated between the mean values of M50 and M57. Effect sizes (*ES*, Cohen, 1988) were calculated for each discrete measure. Cohen's d classification of effect size magnitude was used whereby, d < 0.19 negligible effect; d = 0.20–0.49 small effect; d = 0.50–0.79 moderate effect; d > 0.8 large effect.

RESULTS: The discrete measures for 1st peak vertical and braking GRF significantly increased following a seven-year period of ageing although SPM found no significant differences throughout the stance phase. The normalised knee moment at amortisation significantly increased however SPM revealed no difference except for a significant decrease with ageing during the first 5% of the stance phase. A significant increase in the hip extensor moment occurred during ~25-40% of the stance phase. There were significant differences between the M50 and M57 for step length and a subsequent a decrease in step frequency.

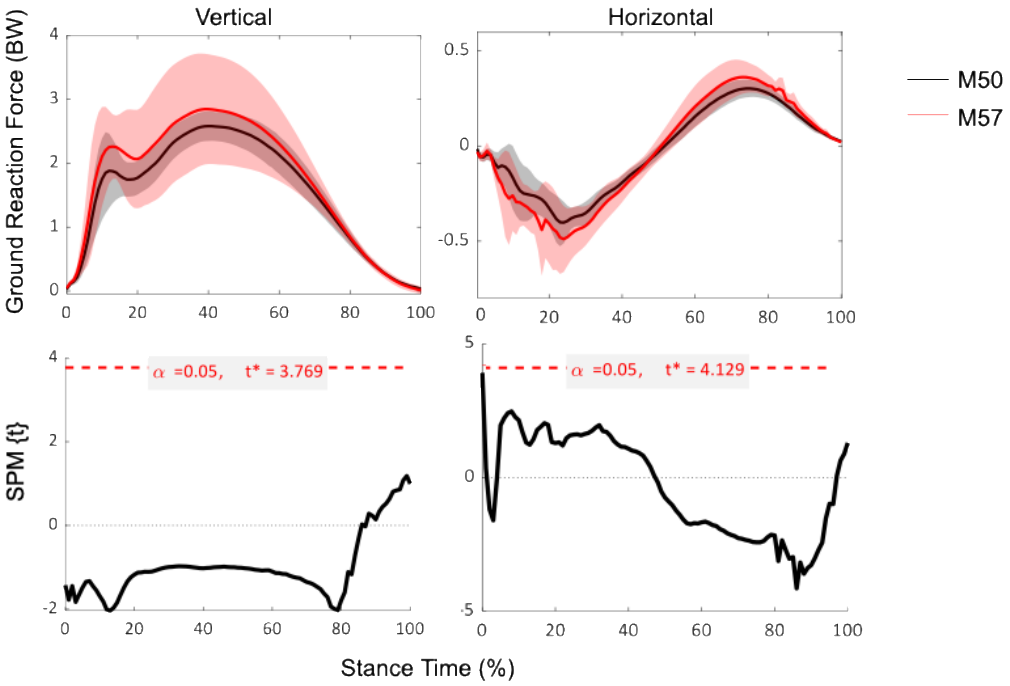

Figure 1: Mean ground reaction force (shaded area = ± 1sd) with the outcomes of the statistical parametric mapping below.

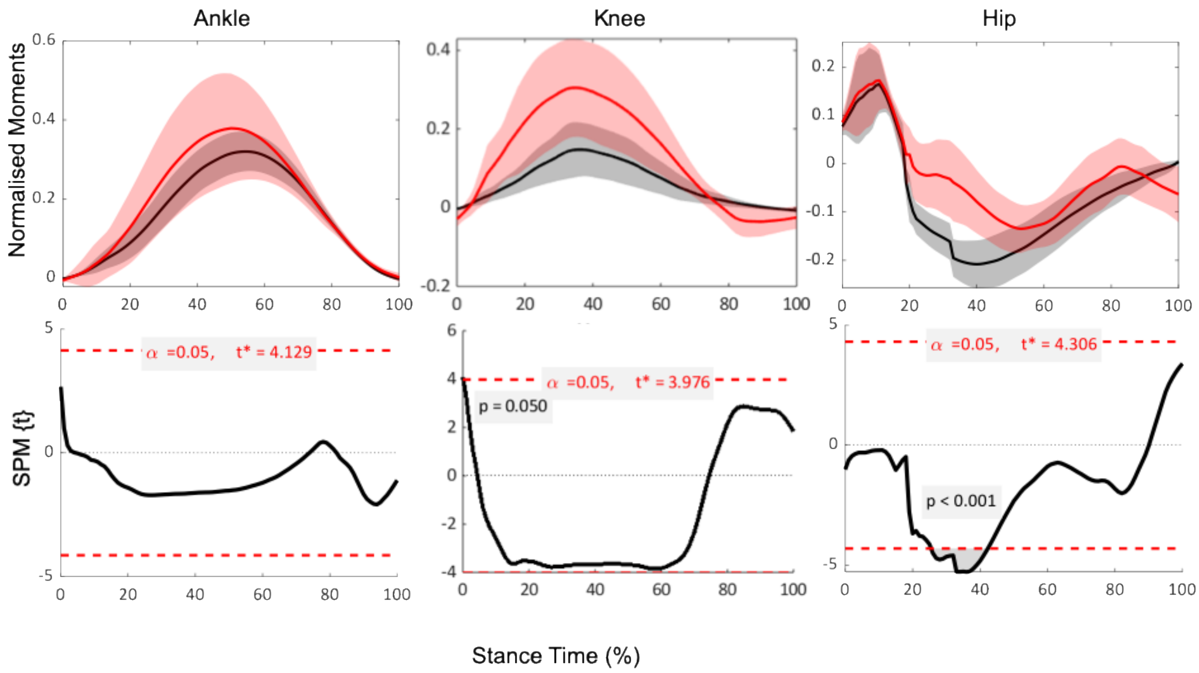

Figure 2: Mean normalised ankle, knee and hip joint moments (shaded area = ± 1sd) with the outcomes of the statistical parametric mapping below.

Table 1: Statistical comparison of the average ±sd discrete measures between M50 and M57.

Measure	M50	M57	*p*	*%diff*	*ES*
Running velocity ($m \cdot s^{-1}$)	3.81±0.39	3.83±0.40	*0.57*	*0.52*	0.05
Step length (m)	1.35±0.21	1.48±0.17	*0.01*	*9.63*	0.65
Step frequency (Hz)	2.81±0.27	2.61±0.28	*0.01*	*7.12*	0.70
1st peak vertical ground reaction force (BW)	1.92±0.38	2.33±0.40	*0.01*	*21.38*	1.01
2nd peak vertical ground reaction force (BW)	2.61±0.25	2.79±0.67	*0.34*	*6.90*	0.34
Peak horizontal braking force (BW)	-0.44±0.12	-0.58±0.19	*0.03*	*31.82*	0.84
Peak horizontal propulsion force (BW)	0.31±0.05	0.37±0.11	*0.07*	*19.35*	0.67
Time of amortisation (%)	45.86±2.95	40.53±2.53	*0.17*	*13.80*	*2.04*
Ankle moment @ amortisation	0.26±0.06	0.24±0.06	*0.40*	*7.69*	0.32
Knee moment @ amortisation	0.12±0.06	0.31±0.07	*0.00*	*158.33*	2.80
Hip moment @ amortisation	0.17±0.06	0.17±0.10	*0.82*	*0.00*	0.00

DISCUSSION: There appears to be two mechanical responses during master athletes' ageing. Firstly, an increased function to attenuate impact forces places the runner under further risk of injury and a mechanism to absorb such forces has seen an increase in lower body joint coordination (Diss et al., 2019). Secondly the change in hip flexor moment (~25-40% of the stance), as the body approaches the shift from a downward to an upward motion at amortisation, places a reliance on the knee extensor towards this transition, which is concerning due to the strength reduction of the triceps surae and quadriceps femoris muscle-tendon units (De Vita et al., 2016) associated with age.

CONCLUSION: Increased tri-joint coordination (Diss et al., 2019) with age are a response mechanism to a significant amplification of the passive impact forces. It is suggested that sequential, coordinated movement patterns are practised with age to minimise injuries that are associated with loading.

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