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PEAK MUSCLE AND JOINT CONTACT FORCES OF RUNNING WITH INCREASED DUTY FACTORS

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(supplementary material from this study can be found in **appendix D**)

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1.1. Abstract

PURPOSE: Running with increased duty factors (DF) has been shown to effectively reduce external forces during running. In this study, we investigated whether running with increased DF (INCR) also reduces internal musculoskeletal loading measures, defined as peak muscle forces, muscle force impulses and peak joint contact forces compared to a runners' preferred running pattern (PREF).

METHODS: Ten subjects were instructed to run with increased DF at 2.1 m•s^{-1}. Ground reaction forces and three-dimensional kinematics were simultaneously measured. A musculoskeletal model was used to estimate muscle forces based on a dynamic optimization approach, which in turn were used to calculate muscle force impulses and (resultant and three-dimensional) joint contact forces of the ankle, knee and hip joint during stance.

RESULTS: Runners successfully increased their DF from 40.6% to 49.2% on average. This reduced peak muscle forces of muscles that contribute to support during running, i.e. the ankle plantar flexors (-19%), knee extensors (-18%) and hip extensors (-15%). As a consequence, peak joint contact forces of the ankle, knee and hip joint reduced in the INCR condition. However, several hip flexors generated higher peak muscle forces near the end of stance.

CONCLUSION: Running with increased DF lowers internal loading measures related to support during stance. Although some swing-related muscles generated higher forces near the end of stance, running with increased DF can be considered as a preventive strategy to reduce the occurrence of running-related injuries, especially in running populations that are prone to overuse injuries.

Keywords: Biomechanics, gait retraining, internal loading, injury, musculoskeletal modelling

1.2. Introduction

Recreational running can be considered as one of the most popular types of physical activity [1,2]. Despite its well-known health benefits [3,4], the incidence of running-related injuries (RRI) is high [5,6], causing runners to interrupt their training program or even abandon running as a leisure activity [7]. The majority of these RRI are overuse injuries which occur when a repetitive load exceeds the structure-specific musculoskeletal capacity, i.e. the load a tissue can withstand before an injury occurs [8,9]. As RRI behave as a mechanical fatigue phenomenon [8], the risk of developing a RRI is primarily determined by the magnitude of peak loading (i.e. loading magnitude), and to a lesser extent by the number of steps taken [8]. This emphasizes the detrimental effect of excessive loading on the musculoskeletal system during running. As such, many runners could benefit from preventive strategies to avoid RRI, especially when these strategies aim to reduce the loading magnitude on the musculoskeletal system.

Running with increased duty factors (DF - ratio of contact time on stride time) has recently been proposed as a strategy to reduce loading magnitude [10,11]. According to Malisoux et al., who investigated injury risk factors in 800+ recreational runners, running with lower DF can be considered as an important risk factor for sustaining RRI [12]. They suggested that a strategy aiming at increasing DF could be effective to prevent overuse injuries in runners. This concurs with spring-mass dynamics and the impulse-momentum theorem, stating that prolonging the stance phase of running, in relation to stride time, results in lower average and peak vertical ground reaction forces (vGRF) [13–15]. In our previous study, subjects successfully altered their running pattern by increasing DF from 42% to 51% on average [10]. As a result, the vertical excursion of the body center-of-mass (bCOM), as well as several loading parameters such as peak vGRF (-20%) and peak eccentric ankle (-34%) and knee power (-26%) reduced significantly. Although these reductions seem promising to promote running with increased DF to an injury-prone population, these parameters are based on external GRFs, which are a surrogate measure for internal loading (e.g. muscle forces) and underestimate the actual loads experienced by the lower limbs [16–19]. These internal forces, which account for e.g. bi-articular muscle activity and individual muscle contributions to the overall musculoskeletal loading, can therefore be considered as a more direct measure of the loads responsible for injuries [16,20]. Hence, in-depth musculoskeletal analyses investigating the internal loading of running with increased DF might shed more light on the possibility of using this running pattern as a viable strategy to reduce loading magnitude and eventually the incidence of RRI.

The main goal of this study was to investigate the effect of running with increased DF on musculoskeletal loading. In this study, musculoskeletal loading was defined as a combination of peak muscle forces, muscle force impulses and (three-dimensional and resultant) peak joint contact forces of the ankle, knee and hip joint. As the greatest peak loads occur during stance (e.g. [16,19,21]), we focused on the stance phase of running. Due to the lower vertical excursion of the bCOM when running with increased DF, less muscle force needs to be delivered to counteract gravity [10]. Therefore we hypothesized that running with increased DF will mainly reduce peak muscle forces of muscles that contribute to support during running, i.e. the main ankle plantar flexors (m. soleus and m. gastrocnemius), knee extensors (m. vasti and m. rectus femoris) and hip extensors (m. gluteus medius and maximus) [22]. It follows that we also expect the ankle, knee and hip peak joint contact forces to be lower. As the stance phase is extended when running with increased DF [10], but lower peak muscle forces can be expected, we hypothesize similar muscle force impulses compared to the runners' preferred running pattern. In general, we expect running with increased DF to be a running pattern which can reduce the 'overall' musculoskeletal loading of the lower limbs, thereby reducing the risk of RRI.

1.3. Methods

Subjects

Ten male subjects (24 $\pm$ 2 year, 71.4 $\pm$ 6.4 kg, 1.83 $\pm$ 0.04 m, BMI: 21.4 $\pm$ 1.6 kg•m^{-2}) who were active in sports participated in this experiment. All subjects were pain- and injury-free for six months and could run continuously for at least one hour. All subjects signed an informed consent before the start of the experiment. The study was approved by the Ethical Committee of the Ghent University hospital.

Experimental design

All runners performed two running conditions with neutral running shoes (Mizuno Wave Rider 18) on a force-instrumented split-belt treadmill (Bertec Corp., Columbus, OH). Before the start of the actual experiment, a habituation protocol was conducted. Subjects walked and ran for 5 min at speeds ranging from 1.4 to 3.2 m•s^{-1} and were instructed to run with higher DF for 5 min at a pace of 2.1 m•s^{-1}. In order to run with higher DF, subjects received the verbal instruction 'try to run without a flight phase' and a short demonstration that was given by the researchers. During the actual experiment, subjects performed two running conditions at 2.1 m•s^{-1} in randomized order. Each condition lasted for 5 min with 3 min of rest in between: running with increased DF (INCR condition) based on the aforementioned verbal instruction and demonstration, and running with their preferred running pattern (PREF). In the latter condition, subjects were asked to run how they 'normally' run during a running session outside.

Measurements

Subjects performed the PREF condition single-belt and the INCR condition split-belt. This allowed us to capture GRFs of both legs separately in the latter condition, in case DF exceeded 50% (i.e. double support phase). When subjects ran split-belt, they were not instructed to place their right foot on the right belt and their left foot on the left belt. Instead, they focused on an extension of the midline between the two belts drawn on the floor in front of the treadmill. Foot contacts that hit both belts simultaneously were deleted from data analysis. GRFs and three-dimensional kinematics (12 Qualisys Oqus cameras, Göteborg, Sweden) were simultaneously measured during the last 10 sec of each condition, at 1000Hz and 250Hz respectively. A full-body kinematic marker set that consisted of 72 reflective markers was used to measure the motion-capture data (see appendix D, Supplemental Digital Content 1, that describes the used marker set).

Data analysis

Spatiotemporal parameters were calculated based on events identified from the vGRF. Timings of initial foot contact and toe-off were determined using a threshold of 20N [23]. DF was calculated as the ratio of contact time on stride time. Contact time was calculated as the time between initial foot contact and toe-off of the same foot, stride time as the time between consecutive initial foot contacts of the same foot and swing time as the time between toe-off and initial foot contact of the same foot. Stride frequency was calculated as the inverse of stride time and stride length was calculated by dividing running speed (i.e. 2.1 $m.s^{-1}$) by stride frequency.

A generic Hamner-model [22] consisting of 29 degrees-of-freedom and 92 musculoskeletal actuators in the lower extremities and torso was used to calculate musculoskeletal loading. This model was scaled in Opensim 3.3 [24] based on the marker positions during a static trial and the subjects' body mass. Maximal isometric muscle forces were also scaled based on the subject's body mass and height [25]. First, marker coordinates and GRF data were low-pass filtered at 15Hz, using a fourth-order Butterworth recursive filter. Peak vGRF and peak braking forces (PBF) were calculated as the maximal value of the vGRF and the minimal value of the anterior-posterior GRF (ant-post GRF) respectively. Next, joint angles and moments were obtained through inverse kinematics and inverse dynamics, implemented in Opensim. As invasive measurements of muscle forces are not possible, an optimization approach was used to solve the muscle redundancy problem. We estimated muscle forces during the stance phase of running in both conditions by using a dynamic optimization algorithm that takes muscle-tendon dynamics into account while minimizing the sum of muscle excitations squared [26]. We preferred a dynamic optimization approach over a static optimization approach as the latter simplifies muscle-tendon dynamics by neglecting activation dynamics and assuming rigid tendons. Inverse dynamic joint moments along with muscle-tendon unit lengths and moment arms were used as inputs to solve the muscle redundancy problem. Dynamic optimization validity was verified 1) by comparing the muscle activation patterns in this study (see appendix D, Supplemental digital content 2, which illustrates muscle activation patterns) with measured (EMG [27,28]) and estimated (modeled [16,21]) muscle activations from literature, and 2) when reserve actuator torques (i.e. torques that are added about each joint to enable the simulation to run when muscles cannot produce the needed force at a given point) were lower than 10% of the respective joint moment, as recommended by Opensim guidelines. The output of the dynamic optimization approach are muscle-tendon forces, but will be referred to as muscle forces in this paper. Peak muscle forces were calculated as the peak value during the stance phase for muscles that contribute most to support during running, which are the main ankle plantar flexors: m. soleus (SOLEUS), m. gastrocnemius medialis (GAS MED) and lateralis (GAS LAT); knee extensors: m. rectus femoris (REC FEM); m. vatus lateralis (VAS LAT) and medialis (VAS MED) and hip extensors: the m. gluteus maximus (GLUT MAX) and medius (GLUT MED) [22]. We also calculated peak muscle forces of the m. psoas (PSOAS), m. iliacus (ILIACUS), m. biceps femoris (BIC FEM), m. semimembranosus (SEMIMEM) and semitendinosus (SEMITEN) as these muscles fulfill an important (swing-related) function during running. The BIC FEM total muscle force was calculated by summing the muscle force of the long head and short head of the BIC FEM. Likewise, the GLUT MAX and GLUT MED total muscle force was calculated as the sum of the three muscle fiber bundles (GLUT MAX: superior, middle and inferior part; GLUT MED: anterior, posterior and middle part) of these muscles, as these were modelled separately. Additionally, muscle force impulses were calculated for the aforementioned muscles by taking the integral of muscle force during stance. Lastly, joint contact forces were calculated using the joint reaction analysis in Opensim [29]. These forces are the forces transferred between consecutive bodies as a result of all loads acting on the model and correspond to the internal loads carried by the joint structures. The joint contact forces were reported in the child (distal) body (i.e. femur, tibia and talus for respectively the hip, knee and ankle joint contact forces) and were calculated in the axial, anterior-posterior (AP) and medio-lateral (ML) direction. Peak joint contact forces were calculated as the maximal value during stance. Negative values indicate forces that are directed downwards for the compressive component (i.e. axial), backward for the anterior-posterior component (i.e. AP) and medial for the medio-lateral component (i.e. ML). Resultant joint contact forces were calculated from these three-dimensional joint contact forces as a measure for the 'overall' contact force acting on the respective joint.

To calculate population means, all measures were first averaged over consecutive stance phases of the same leg, then averaged between left and right and then averaged across subjects. On average, 15 and 17 contacts were analyzed for the INCR and the PREF condition respectively for every subject. Figures representing muscle forces and joint contact forces are expressed over averaged contact time (in ms) for the respective condition.

Statistics

Paired samples t-test were performed to compare differences in spatiotemporal characteristics, external force measures, peak muscle forces, muscle force impulses and joint contact forces between the PREF and the INCR condition. The p-value was set at 0.05.

1.4. Results

Subjects increased their DF by 8.6% on average (Table 1, Figure 1). In the INCR condition, contact time and stride frequency increased, while stride time, stride length, swing time, peak vGRF and PBF decreased compared to the PREF condition (Table 1). The vGRF and ant-post GRF are presented in Figure 2.

Table 1: spatiotemporal and external force parameters (MEAN $\pm$ STDEV) for PREF and INCR.

	PREF [MEAN $\pm$ STDEV]	INCR [MEAN $\pm$ STDEV]	[t- ; p-value]
Speed (m.s^{-1})	2.10	2.10	N.A.
Duty factor (%) ↑	40.6 $\pm$ 4.4	49.2 $\pm$ 2.7	[-5.558 ; <0.001]
Contact time (s) ↑	0.32 $\pm$ 0.03	0.37 $\pm$ 0.03	[-4.038 ; 0.003]
Stride time (s) ↓	0.78 $\pm$ 0.03	0.75 $\pm$ 0.03	[4.514; 0.001]
Stride frequency (strides.s^{-1}) ↑	1.28 $\pm$ 0.05	1.33 $\pm$ 0.05	[-4.221; 0.002]
Stride length (m) ↓	1.65 $\pm$ 0.06	1.58 $\pm$ 0.06	[4.323; 0.002]
Swing time (s) ↓	0.47 $\pm$ 0.04	0.38 $\pm$ 0.02	[5.937 ; <0.001]
Peak vGRF (BW) ↓	2.32 $\pm$ 0.25	1.92 $\pm$ 0.16	[4.722 ; 0.001]
PBF (BW) ↓	-0.24 $\pm$ 0.02	-0.21 $\pm$ 0.03	[-3.38 ; 0.008]

INCR: increased condition; PREF: preferred condition; peak vGRF: peak vertical ground reaction force; PBF: peak braking force; N.A.: not applicable. As speed was equal between conditions, no standard deviation was given. ↓ indicates INCR < PREF; ↑ indicates INCR > PREF (with $p < 0.05$).

All muscle forces are depicted in Figure 3. Peak muscle forces during stance (Table 2) were significantly lower in the INCR condition compared to the PREF condition for the following muscles: SOLEUS, GAS MED, GAS LAT, VAS LAT, VAS MED, SEMITEN, BIC FEM, GLUT MED and GLUT MAX. Peak muscle forces were higher in the INCR compared to the PREF condition for the PSOAS, ILIACUS, and SEMIMEM. Muscle force impulses (Table 3) of the SOLEUS, GAS MED, GAS LAT and GLUT MED were lower for the INCR compared to the PREF condition, whereas the muscle force impulses of the PSOAS, ILIACUS and SEMIMEM were higher.

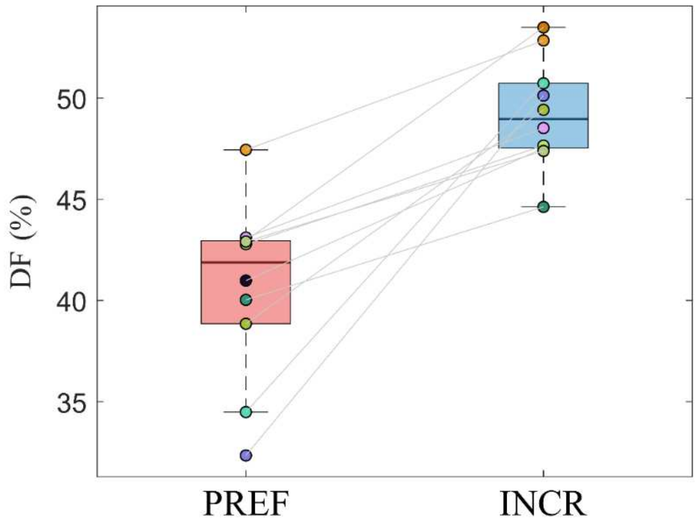

Figure 1: duty factor (DF) for the increased (INCR - blue) and preferred (PREF - red) condition. Boxplot with 25th and 75th percentile, median and outlier (data point outside PREF boxplot; the outlier was not deleted as it represents a true value and was not due to an artefact) representing DF for the PREF and INCR condition. Data points are averaged DF for all (n = 10) individuals and are connected between conditions for the respective subject (grey lines). Data points are color coded and are unique for every subject.

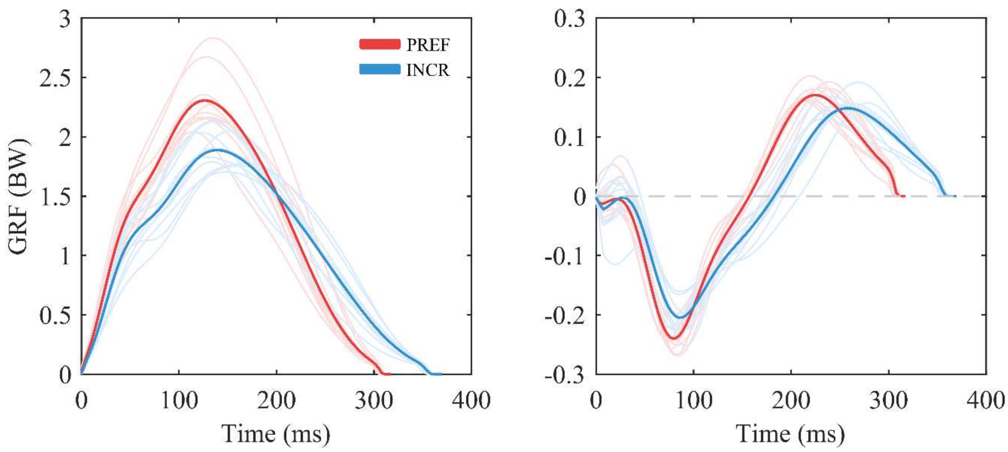

Figure 2: Ensemble averages of the vertical (left) and anterior-posterior (right) ground reaction forces (GRF) of the increased (INCR - blue) and preferred (PREF - red) condition. Forces were averaged over consecutive steps and subjects (n = 10) during stance (expressed as averaged contact time (ms) for the respective condition). Transparent lines represent the individual (averaged) ground reaction force of each subject.

All peak joint contact forces (Table 4) were significantly lower for the INCR condition compared to the PREF condition, except for the ML component in the ankle joint where no difference was found. Figure 4 shows the resultant joint contact force for the ankle, knee and hip joint. A figure of the joint contact forces in the axial, AP and ML direction of the respective joints can be found as supplementary material (see appendix D, Supplemental digital content 3, which illustrates joint contact forces in all directions).

Table 2: peak muscle forces (MEAN ± STDEV) for PREF and INCR.

Peak muscle forces (BW)	**PREF** [MEAN ± STDEV]	**INCR** [MEAN ± STDEV]	[t- ; p-value]
SOLEUS ↓	5.80 ± 1.03	4.71 ± 0.72	[5.264 ; 0.001]
GAS MED ↓	1.07 ± 0.12	0.92 ± 0.08	[5.438 ; <0.001]
GAS LAT ↓	0.35 ± 0.05	0.29 ± 0.03	[4.950 ; 0.001]
VAS LAT ↓	2.10 ± 0.35	1.73 ± 0.19	[4.776 ; 0.001]
VAS MED ↓	0.95 ± 0.16	0.79 ± 0.09	[4.723 ; 0.001]
REC FEM	0.91 ± 0.45	0.85 ± 0.28	[1.103 ; 0.299]
BIC FEM ↓	0.75 ± 0.17	0.64 ± 0.17	[3.716 ; 0.005]
SEMIMEM ↑	0.65 ± 0.10	0.81 ± 0.10	[-3.771 ; 0.004]
SEMITEN ↓	0.23 ± 0.07	0.15 ± 0.08	[5.237 ; 0.001]
PSOAS ↑	0.79 ± 0.20	1.02 ± 0.17	[-3.157 ; 0.004]
ILIACUS ↑	0.71 ± 0.17	0.88 ± 0.12	[-2.405 ; 0.040]
GLUT MED ↓	3.37 ± 0.35	2.87 ± 0.21	[4.986 ; 0.001]
GLUT MAX ↓	0.97 ± 0.27	0.83 ± 0.21	[2.656; 0.026]

INCR: increased condition; PREF: preferred condition; Abbreviation muscles: m. soleus (SOLEUS), gastrocnemius medialis and lateralis (GAS MED and GAS LAT), vastus medialis and lateralis (VAS MED and VAS LAT), rectus femoris (REC FEM), biceps femoris (BIC FEM), semimembranosus (SEMIMEM), semitendinosus (SEMITEN), psoas (PSOAS), iliacus (ILIACUS), gluteus medius and maximus (GLUT MED and GLUT MAX). ↓ indicates INCR < PREF; ↑ indicates INCR > PREF (with $p < 0.05$).

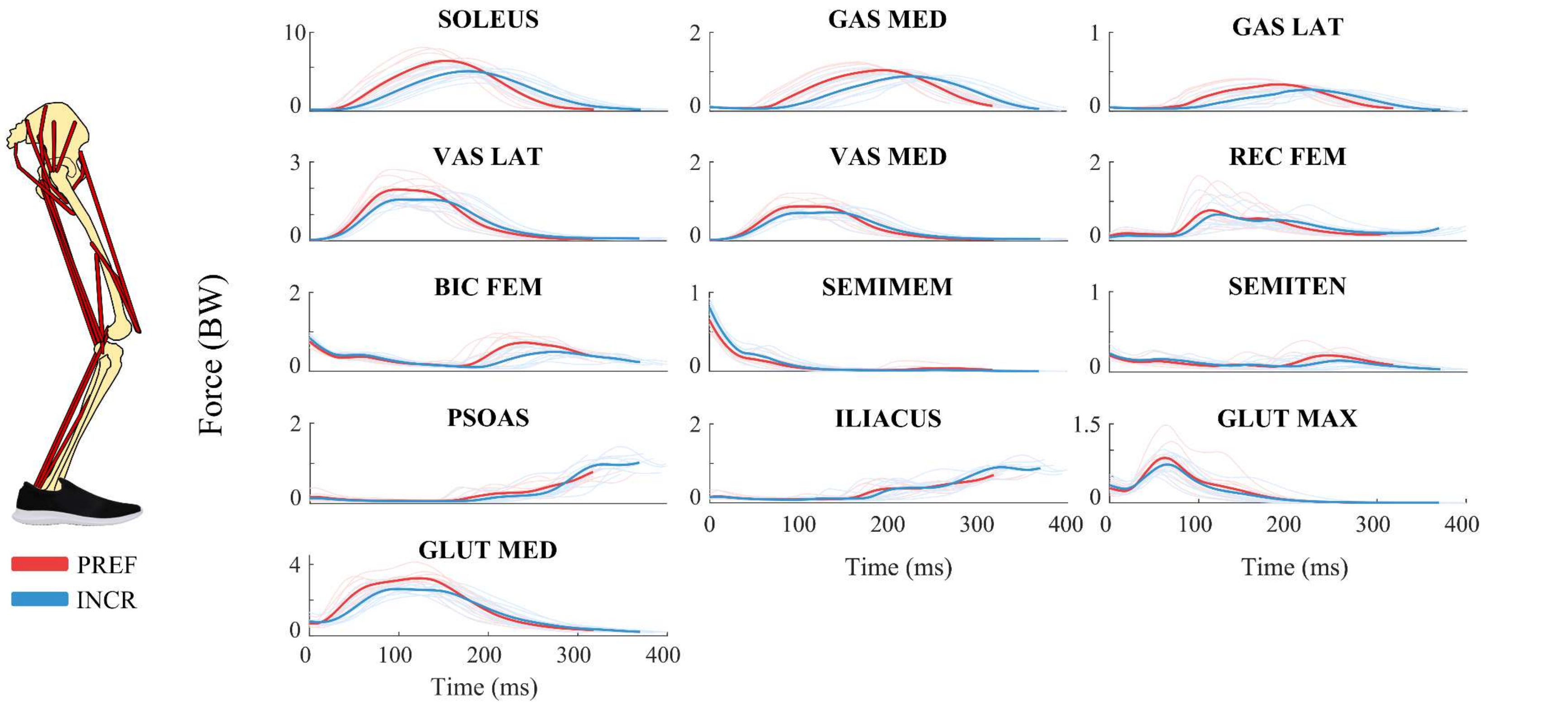

Figure 3: Ensemble averages of muscle forces for the increased (INCR - blue) and preferred (PREF - red) condition. Muscle forces were averaged over consecutive steps and subjects (n = 10) during stance (expressed as averaged contact time (ms) for the respective condition). Transparent lines represent the individual (averaged) muscle forces of each subject. All muscles included in this figure are presented on the figure of the model (left of graph). Figure of the musculoskeletal model was used by permission (Hamner SR, Seth A, Delp SL. Muscle contributions to propulsion and support during running. *J Biomech*. 2010;43(14):2709–16).

Table 3: muscle force impulses (MEAN ± STDEV) for PREF and INCR.

Muscle force Impulses (BWs)	**PREF** MEAN-STDEV	**INCR** MEAN-STDEV	[t- ; p-value]
SOLEUS ↓	0.82 ± 0.10	0.74 ± 0.09	[5.295 ; 0.001]
GAS MED ↓	0.16 ± 0.02	0.15 ± 0.02	[5.084 ; 0.001]
GAS LAT ↓	0.05 ± 0.01	0.04 ± 0.01	[5.268 ; 0.001]
VAS LAT	0.25 ± 0.05	0.24 ± 0.04	[1.518 ; 0.163]
VAS MED	0.11 ± 0.02	0.11 ± 0.02	[1.016 ; 0.336]
REC FEM	0.11 ± 0.04	0.12 ± 0.02	[-1.311 ; 0.222]
BIC FEM	0.13 ± 0.02	0.12 ± 0.02	[1.686 ; 0.126]
SEMIMEM ↑	0.02 ± 0.01	0.03 ± 0.01	[-4.606 ; 0.001]
SEMITEN	0.04 ± 0.01	0.04 ± 0.01	[0.020 ; 0.984]
PSOAS ↑	0.07 ± 0.01	0.11 ± 0.02	[-4.904 ; 0.001]
ILIACUS ↑	0.08 ± 0.01	0.13 ± 0.02	[-5.818 ; <0.001]
GLUT MED ↓	0.54 ± 0.05	0.49 ± 0.04	[3.753 ; 0.005]
GLUT MAX	0.08 ± 0.03	0.07 ± 0.03	[1.260 ; 0.239]

INCR: increased condition; PREF: preferred condition; Abbreviation muscles: m. soleus (SOLEUS), gastrocnemius medialis and lateralis (GAS MED and GAS LAT), vastus medialis and lateralis (VAS MED and VAS LAT), rectus femoris (REC FEM), biceps femoris (BIC FEM), semimembranosus (SEMIMEM), semitendinosus (SEMITEN), psoas (PSOAS), iliacus (ILIACUS), gluteus medius and maximus (GLUT MED and GLUT MAX). ↓ indicates INCR < PREF; ↑ indicates INCR > PREF (with $p < 0.05$).

1.5. Discussion

The aim of this study was to investigate whether running with increased DF could reduce the internal forces acting on the musculoskeletal system, defined as peak muscle forces, muscle force impulses and joint contact forces. Overall, this study showed that running with increased DF reduced peak muscle forces of muscles with a support-related function during stance, and as a consequence peak joint contact forces of the ankle, knee and hip joint. Besides these reductions, several hip flexor muscles related to the swing-phase of running generated higher peak muscle forces near the end of stance in the INCR condition compared to the PREF condition.

Table 4: Peak ankle, knee and hip joint contact forces (MEAN ± STDEV) for PREF and INCR.

Peak joint contact forces (BW)	**PREF** [MEAN ± STDEV]	**INCR** [MEAN ± STDEV]	[t- ; p-value]
Ankle AP ↓	3.34 ± 1.02	2.74 ± 0.69	[4.435 0.002]
Ankle Axial ↓	-9.43 ± 1.43	-7.29 ± 0.99	[-5.948 <0.001]
Ankle ML	-0.10 ± 0.05	-0.09 ± 0.03	[-0.346 0.737]
Ankle resultant ↓	**10.01 ± 1.65**	**7.79 ± 1.12**	**[5.958 ; <0.001]**
Knee AP ↓	-4.45 ± 0.73	-3.53 ± 0.45	[-6.760 <0.001]
Knee Axial ↓	-6.78 ± 0.85	-5.44 ± 0.65	[6.177 <0.001]
Knee ML ↓	-0.39 ± 0.07	-0.33 ± 0.05	[-3.463 0.007]
Knee resultant ↓	**8.10 ± 1.03**	**6.37 ± 0.59**	**[6.394 ; <0.001]**
Hip AP ↓	2.23 ± 0.36	1.94 ± 0.20	[3.634 0.005]
Hip Axial ↓	-6.73 ± 0.70	-5.45 ± 0.47	[-7.018 <0.001]
Hip ML ↓	2.43 ± 0.37	1.81 ± 0.20	[7.217 <0.001]
Hip resultant ↓	**7.49 ± 0.76**	**5.96 ± 0.42**	**[7.826 ; < 0.001]**

INCR: increased condition; PREF: preferred condition; AP: antero-posterior; ML: medio-lateral; ↓ indicates INCR < PREF; ↑ indicates INCR > PREF (with $p < 0.05$);

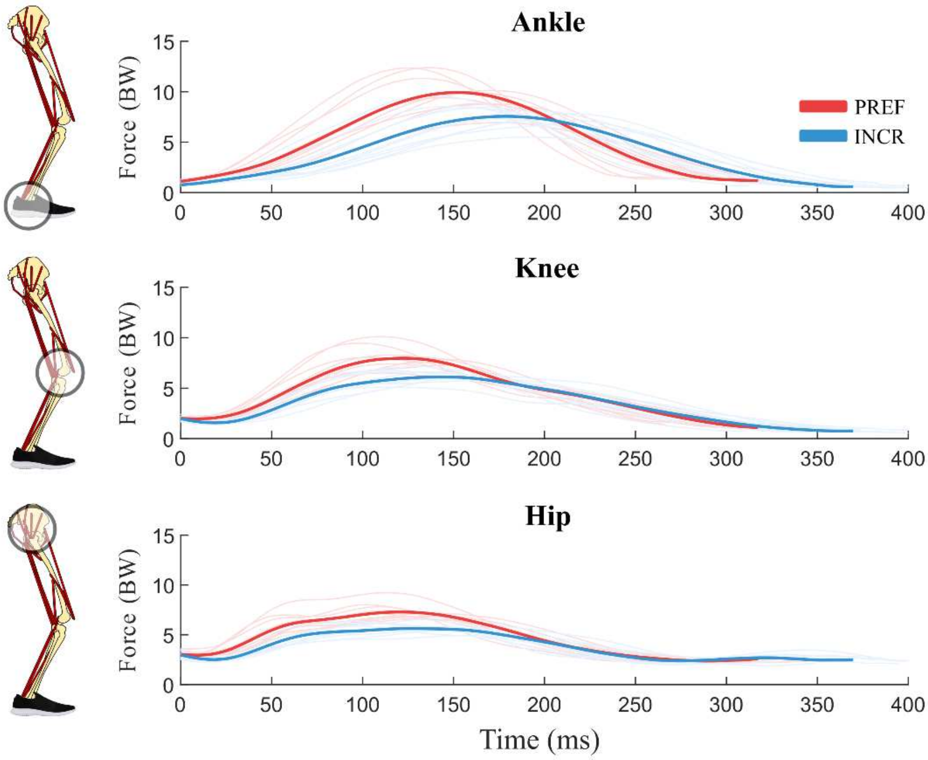

Figure 4: Ensemble averages of resultant joint contact forces for the increased (INCR - blue) and preferred (PREF - red) condition. Resultant joint contact forces were averaged over consecutive steps and subjects (n = 10) during stance (expressed as averaged contact time (ms) for the respective condition). Transparent lines represent the individual (averaged) resultant joint contact forces of each subject. Figure of the musculoskeletal model (left of graph) was used by permission (Hamner SR, Seth A, Delp SL. Muscle contributions to propulsion and support during running. *J Biomech*. 2010;43(14):2709–16).

Subjects increased their DF by increasing contact time (+16%) and decreasing swing time (-19%). As stride time decreased (-4%) and speed was kept similar between conditions, stride length and stride frequency decreased and increased respectively. As a result, peak vGRF and peak muscle forces of the muscles that contribute to support during running reduced significantly: peak muscle forces of the main ankle plantar flexors (SOLEUS, GAS MED and GAS LAT) reduced up to 19%, of the knee extensors (VAS MED and VAS LAT) up to 18% and of the hip extensors (GLUT MED and GLUT MAX) up to 15%. Also the muscle force impulses of the main ankle plantar flexors reduced up to 20%. These results concur with findings of Dorn et al. [30], who related reductions in peak vGRF to lower peak muscle forces of SOLEUS, GAS and VAS. As these muscles provide about 75% of the total vertical support impulse needed to accelerate the body upward, the reductions in peak muscle forces in the current study indicate that running with increased DF engages the muscles that are involved in body weight support less intensely. Furthermore, these reductions resulted in lower ankle, knee and hip peak joint contact forces (up to 26%). Although these reductions are promising in terms of RRI prevention, transferring them to a lower risk/incidence of RRI is not straightforward because the aetiology of RRI is multifactorial and does not only depend on the structure-specific musculoskeletal loading, but also on the runners' musculoskeletal capacity [9]. However, literature does suggest a relationship between increased internal load measures and several RRI. For example: increased joint contact forces in the ankle joint could relate to tibial stress fractures [31], increased joint contact forces in the knee joint to the occurrence of knee osteoarthritis [32] and increased loading of the Achilles tendon to Achilles tendon injuries [33].

Besides the increase in DF, stride frequency and stride length also changed when running with increased DF, which could have influenced musculoskeletal loading as well. According to previous literature [34–37], these spatiotemporal adjustments can also reduce peak vGRF, extensor muscle forces and joint contact forces similar to the reductions found in this study. As changes in stride frequency and stride length can go together with changes in DF and vice versa (e.g. 14, 36), it is difficult to attribute reductions in musculoskeletal loading to changes in one specific spatiotemporal variable (e.g. DF), independent of changes in other spatiotemporal variables (e.g. stride frequency). However, findings of Morin et al. [14] suggest that stride frequency rather indirectly affects spring-mass characteristics through its effect on contact time in relation to stride time, and thus DF. Nonetheless, which spatiotemporal adjustments mainly contribute to reductions in musculoskeletal loading remains an open question. As such, future gait-retraining studies should specifically focus on the relationship between various spatiotemporal characteristics, their interaction with each other, and musculoskeletal loading. Such studies can provide fundamental insights in the spatiotemporal determinants of load, which can guide the development of effective gait retraining strategies to reduce RRI. Additionally, the increase in stride frequency found in this study also indicates a higher repetition of the forces acting on the musculoskeletal system. However, as the risk of developing a RRI is determined to a greater extent by peak loading than by the numbers of steps taken [8], it can be argued that running with increased DF can be considered as a promising gait-retraining strategy with the aim of avoiding RRI, while maintaining a certain minimal load necessary to keep for example tissue remodeling ongoing [39].

One of the most common overuse injuries in running is patellofemoral (PF) pain. In this study we could not determine PF forces because the Hamner model does not contain a PF joint [22]. However, the results of this study contain kinematics and muscle forces which can be used to get a rough estimate of the PF forces, which in turn are related to PF pain. PF compressive forces have been positively related to both muscle forces of the m. quadriceps, peak stance phase knee flexion and peak vGRF [34,40]. In this study, lower muscle forces of the m. quadriceps (VAS MED: -17%; VAS LAT: -18%) were found in the INCR condition. This implies, in combination with a lower peak vGRF (-17%) and no difference in maximal knee flexion (see appendix D, Supplemental digital content 4, which illustrates sagittal joint angles during stance), that running with increased DF will most likely result in lower PF forces.

Whereas running with increased DF reduces peak muscle forces of the main muscles contributing to the support phase of running, several swing-related muscles generated higher forces in the INCR condition near the end of stance. Subjects increased their DF by increasing contact time and decreasing stride time, while stride frequency increased slightly. These spatiotemporal adjustments resulted in less time to swing the legs forward, which might explain the higher peak muscle forces (up to 25%, see Figure 3) of the hip flexors near the end of stance (i.e. PSOAS, ILIACUS). Although the hip flexors generated higher peak muscle forces in the INCR condition compared to the PREF condition, the resultant hip joint contact force was lower. This can be attributed to the temporal difference between the occurrence of the peak resultant hip joint contact force (near mid-stance) and the occurrence of the peak hip flexor muscle forces (near the end of stance). Additionally, at the timing of peak resultant hip joint contact force, peak hip flexor muscle forces (up to 0.09 BW) were much lower compared to peak muscle forces of support-related muscles located at the hip (up to 3.27 BW), indicating that the hip flexor muscles did not substantially contribute to the peak joint contact force of the hip joint. However, since we did not incorporate the flight phase in our analyses, it is difficult to make conclusive statements regarding swing-related muscles.

Several strategies arise to achieve a reduction in the incidence of RRI. A first strategy is to focus on a specific part of the running population that is at higher risk of RRI, such as novice runners [41]. As the incidence of RRI is high in these runners, it can be argued that their musculoskeletal system is less familiar with the acting repetitive forces, which makes them more susceptible for overuse injuries. Running with increased DF could then serve as an optimal 'entry point' to start with progressive running programs (in terms of loading) for these runners, as it decreases the 'overall' loading compared to their preferred running pattern. Also runners who suffered from a (recurring) injury, but are ready to return to running, might benefit from this adjustment in running style to gradually built up the experienced loading. Another strategy is to target runners that suffer from specific injuries which are related to the reductions found in this study, such as tibial stress fractures [31], knee osteoarthritis [32] and Achilles tendon injuries [33]. It holds for both strategies that, given its ease of execution (especially at slow speeds), this gait-retraining strategy could be implemented in intervention studies to investigate its effectiveness on injury incidence. However, as mentioned before, extra care should be given to runners who suffer from injuries located around the hip joint.

Although the calculated muscle forces and joint contact forces correspond well with those reported in literature, it is not straightforward to compare them with other studies due to differences in performed analyses, used models and running speed [16]. In this study, the internal force measures are somewhat lower compared to literature, which is probably due to the difference in speed. While our subjects ran at a rather slow pace of 2.1 $m \cdot s^{-1}$, almost all other studies calculating internal force measures were performed at faster running speeds (3.2 to 5.3 $m \cdot s^{-1}$). Nonetheless, the simulated muscle activation patterns (see appendix D, Supplemental digital content 2, which illustrates muscle activations patterns) correspond well with those estimated in other modelling studies (e.g. [16,17,44–47,19–21,32,34,35,42,43]), even when taking methodological differences into account. Additionally, the muscle activation patterns in this study agree well with EMG data from other studies [27,28], which contributes to the validity of the model estimates.

While modelling studies can aid in finding the right modalities to implement running with increased DF in substantiated training programs, several aspects remain to be elucidated. Therefore, future studies should 1) investigate the effect of running with increased DF and stride frequencies on musculoskeletal loading, independent of one another, 2) focus on the dose-response relationship between DF and internal load measures, 3) investigate whether running with increased DF can be learned and maintained over a longer period and 4) investigate the relationship between running with increased DF and injury incidence in large-scaled intervention studies. It should also be noted that this study has several limitations. Foremost, the experiment was performed on healthy subjects at a slow running speed and not on a population that might benefit most from this gait retraining strategy (e.g. the novice runner that is often injured). Nonetheless, there are no reasons to believe that the mechanics behind the reductions in internal load measures will be different when applied to other populations, mainly due to the strong link between DF and spring-mass dynamics [14]. Secondly, in the INCR condition runners ran split-belt which might have influenced their gait characteristics. However, in order to calculate the documented loading parameters when running with DF > 50%, GRFs needed to be captured for both legs separately. Additionally, we did not include the swing phase in our analyses, as we were mainly interested in peak muscle and joint contact forces which occur during stance. To gain more insight into possible adverse effects of running with increased DF, we suggest incorporating the swing phase in future studies. Finally, the muscle redundancy problem was solved by minimizing muscle excitations squared. This implies a minimization of co-contraction and thus represents a 'most optimal' scenario when calculating internal loading parameters. As such, our findings do not entirely reflect the reality of muscle behavior, especially in the INCR condition as runners changed their running pattern towards a running style they were not familiar with. However, the results of this study indicate how a reduction in musculoskeletal loading of certain anatomical structures can be achieved once a runner is more adapted to this running style.

In conclusion, running with increased duty factors reduced peak muscle forces of muscles which contribute to support during running: i.e. the ankle plantar flexor muscles (m. gastrocnemius and soleus), knee extensor muscles (m. vastus lateralis and medialis) and hip extensor muscles (m. gluteus maximus and medius). Consequently, this resulted in lower peak joint contact forces in the ankle, knee and hip joint. Due to a higher stride frequency and a shorter swing time when running with increased duty factors, several hip flexors (i.e. m. psoas and iliacus) generated higher muscle forces near the end of stance, suggesting an increased loading of swing-related muscles during the swing phase. These results show that running with increased DF can be considered as a potential gait-retraining strategy to avoid overuse injuries, mainly in a population that is prone to running-related injuries.

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Conflict of interest and Source of Funding

All authors declare no conflict of interest. The results of the study are presented clearly, honestly and without fabrication, falsification, or inappropriate data manipulation. The results of the present study do not constitute endorsement by ACSM.

1.6. References

1. Runrepeat Web site [Internet]. Fort Collins (CO): Runrepeat; [cited 2021 Sep 21]. Available from: https://runrepeat.com/state-of-running

2. Tudor-Locke C, Johnson WD, Katzmarzyk PT. Frequently Reported Activities by Intensity for U.S. Adults. *Am J Prev Med*. 2010;39(4):13–20.

3. Lee D, Brellenthin AG, Thompson PD, Sui X, Lee I-M, Lavie CJ. Running as a Key Lifestyle Medicine for Longevity. *Prog Cardiovasc Dis*. 2017;60(1):45–55.

4. Pedisic Z, Shrestha N, Kovalchik S, Stamatakis E, Liangruenrom N, Grgic J. Is running associated with a lower risk of all-cause, cardiovascular and cancer mortality, and is the more the better? A systematic review and meta-analysis. *Br J Sports Med*. 2020;54:898-905.

5. Van Gent RN, Siem D, Van Middeloop M, Van Os AG, Bierma-Zeinstra SMA, Koes BW. Incidence and determinants of lower extremity running injuries in long distance runners: A systematic review. *Br J Sports Med*. 2007;40(4):16–29.

6. Lopes AD, Hespanhol LC, Yeung SS, Costa LOP. What are the Main Running-Related Musculoskeletal Injuries? *Sport Med*. 2012;42(10):891–905.

7. Videbæk S, Bueno AM, Nielsen RO, Rasmussen S. Incidence of Running-Related Injuries Per 1000 h of running in Different Types of Runners: A Systematic Review and Meta-Analysis. *Sport Med*. 2015;45(7):1017–26.

8. Edwards WB. Modeling Overuse Injuries in Sport as a Mechanical Fatigue Phenomenon. *Exerc Sport Sci Rev*. 2018;46(4):224–31.

9. Bertelsen ML, Hulme A, Petersen J, Brund RK, Sørensen H, Finch CF. A framework for the etiology of running-related injuries. *Scand J Med Sci Sports*. 2017;27(11):1170–80.

10. Bonnaerens S, Fiers P, Galle S, et al. Grounded Running Reduces Musculoskeletal Loading. *Med Sci Sport Exerc*. 2019;51(4):708–15.

11. Bonnaerens S, Fiers P, Galle S, et al. Relationship between duty factor and external forces in slow recreational runners. *BMJ Open Sport Exerc Med* [Internet]. 2021 [cited 2021 Feb 18]; 7(1). Available from: https://bmjopensem.bmj.com/lookup/doi/10.1136/bmjsem-2020-000996. doi:10.1136/ bmjsem-2020-000996

12. Malisoux L, Gette P, Delattre N, et al. Spatiotemporal and Ground-Reaction Force Characteristics as Risk Factors for Running-Related Injury: A Secondary Analysis of a Randomized Trial Including 800+ Recreational Runners. *Am J Sports Med*. 2022;50(2):537-544

13. Weyand PG, Sternlight DB, Bellizzi MJ, Wright S. Faster top running speeds are achieved with greater ground forces not more rapid leg movements. *J Appl Physiol*. 2000;89(5):1991–9.

14. Morin JB, Samozino P, Zameziati K, Belli A. Effects of altered stride frequency and contact time on leg-spring behavior in human running. *J Biomech*. 2007;40(15):3341–8.

15. Clark KP, Weyand PG. Are running speeds maximized with simple-spring stance mechanics? *J Appl Physiol*. 2014;15;117(6):604–15.

16. Rooney BD, Derrick TR. Joint contact loading in forefoot and rearfoot strike patterns during running. *J Biomech*. 2013;46(13):2201–6.

17. van den Bogert AJ. Analysis and simulation of mechanical loads on the human musculoskeletal system: a methodological overview. *Exerc Sport Sci Rev*. 1994;22(403):23–51.

18. Erdemir A, McLean S, Herzog W, van den Bogert AJ. Model-based estimation of muscle forces exerted during movements. *Clin Biomech*. 2007;22(2):131–54.

19. Scott SH, Winter DA. Internal forces of chronic running injury sites. *Med Sci Sports Exerc*. 1990;22(3):357–69.

20. Meardon SA, Willson JD, Kernozek TW, Duerst AH, Derrick TR. Shoe cushioning affects lower extremity joint contact forces during running. *Footwear Sci*. 2018;10(2):109–17.

21. Thomas JM, Edwards WB, Derrick TR. Joint Contact Forces with Changes in Running Stride Length and Midsole Stiffness. *J Sci Sport Exerc*. 2020;2(1):69–76.

22. Hamner SR, Seth A, Delp SL. Muscle contributions to propulsion and support during running. *J Biomech*. 2010;43(14):2709–16.

23. Watt JR, Franz JR, Jackson K, Dicharry J, Riley PO, Kerrigan DC. A three-dimensional kinematic and kinetic comparison of overground and treadmill walking in healthy elderly subjects. *Clin Biomech*. 2010;25(5):444–9.

24. Delp SL, Anderson FC, Arnold AS, et al. OpenSim: Open-Source Software to Create and Analyze Dynamic Simulations of Movement. *IEEE Trans Biomed Eng*. 2007;54(11):1940–50.

25. Handsfield GG, Meyer CH, Hart JM, Abel MF, Blemker SS. Relationships of 35 lower limb muscles to height and body mass quantified using MRI. *J Biomech*. 2014;47(3):631–8.

26. De Groote F, Kinney AL, Rao AV, Fregly BJ. Evaluation of Direct Collocation Optimal Control Problem Formulations for Solving the Muscle Redundancy Problem. *Ann Biomed Eng*. 2016;44(10):2922–36.

27. Chumanov ES, Wille CM, Michalski MP, Heiderscheit BC. Changes in muscle activation patterns when running step rate is increased. *Gait Posture*. 2012;36(2):231–5.

28. Gazendam MGJ, Hof AL. Averaged EMG profiles in jogging and running at different speeds. *Gait Posture*. 2007;25(4):604–14.

29. Steele KM, DeMers MS, Schwartz MH, Delp SL. Compressive tibiofemoral force during crouch gait. *Gait Posture*. 2012;35(4):556–60.

30. Dorn TW, Schache AG, Pandy MG. Muscular strategy shift in human running: dependence of running speed on hip and ankle muscle performance. *J Exp Biol*. 2012;215(11):1944–56.

31. Sasimontonkul S, Bay BK, Pavol MJ. Bone contact forces on the distal tibia during the stance phase of running. *J Biomech*. 2007;40(15):3503–9.

32. Willy RW, Meardon SA, Schmidt A, Blaylock NR, Hadding SA, Willson JD. Changes in tibiofemoral contact forces during running in response to in-field gait retraining. *J Sports Sci*. 2016;34(17):1602–11.

33. Almonroeder T, Willson JD, Kernozek TW. The Effect of Foot Strike Pattern on Achilles Tendon Load During Running. *Ann Biomed Eng*. 2013;41(8):1758–66.

34. Lenhart RL, Thelen DG, Wille CM, et al. Increasing running step rate reduces patellofemoral joint forces. *Med Sci Sports Exerc.* 2014;46(3):557–64.

35. Lenhart R, Thelen D, Heiderscheit B. Hip Muscle Loads During Running at Various Step Rates. *J Orthop Sport Phys Ther*. 2014;44(10):766-A4.

36. Edwards WB, Taylor D, Rudolphi JT, Gilette JC, Derrick TR. Effects of Stride Length and Running Mileage on a Probabilistic Stress Fracture Model. *Med Sci Sport Exerc.* 2009;41(12):2177–84.

37. Heiderscheit BC, Chumanov ES, Michalski MP, Wille CM, Ryan MB. Effects of Step Rate Manipulation on Joint Mechanics during Running. *Med Sci Sport Exerc*. 2011;43(2):296–302.

38. Adams D, Pozzi F, Willy RW, Carrol A, Zeni J. Altering Cadence or Vertical Oscillation During Running: Effects on Running Related Injury Factors. Int J Sports Phys Ther. 2018;13(4):633–42.

39. Kohrt WM, Bloomfield SA, Little KD, Nelson ME, Yingling VR. Physical Activity and Bone Health. *Med Sci Sport Exerc*. 2004;36(11):1985–96.

40. Yamaguchi GT, Zajac FE. A planar model of the knee joint to characterize the knee extensor mechanism. J Biomech. 1989;22(1):1–10.

41. Kluitenberg B, van Middelkoop M, Diercks R, van der Worp H. What are the Differences in Injury Proportions Between Different Populations of Runners? A Systematic Review and Meta-Analysis. *Sport Med*. 2015;45(8):1143–61.

42. Glitsch U, Baumann W. The three-dimensional determination of internal loads in the lower extremity. *J Biomech*. 1997;30(11–12):1123–31.

43. Edwards WB, Gillette JC, Thomas JM, Derrick TR. Internal femoral forces and moments during running: Implications for stress fracture development. *Clin Biomech*. 2008;23(10):1269–78.

44. Edwards WB, Taylor D, Rudolphi TJ, Gillette JC, Derrick TR. Effects of running speed on a probabilistic stress fracture model. Clin Biomech. 2010;25(4):372–7.

45. Willy RW, Bigelow MA, Kolesar A, Willson JD, Thomas JS. Knee contact forces and lower extremity support moments during running in young individuals post-partial meniscectomy. *Knee Surgery, Sport Traumatol Arthrosc*. 2017;25(1):115–22.

46. Pandy MG, Andriacchi TP. Muscle and joint function in human locomotion. Vol. 12, Annual Review of Biomedical Engineering. 2010; 401–433 p.

47. Bergmann G, Graichen F, Rohlmann A. Hip joint loading during walking and running, measured in two patients. *J Biomech*. 1993;26(8):969–90.