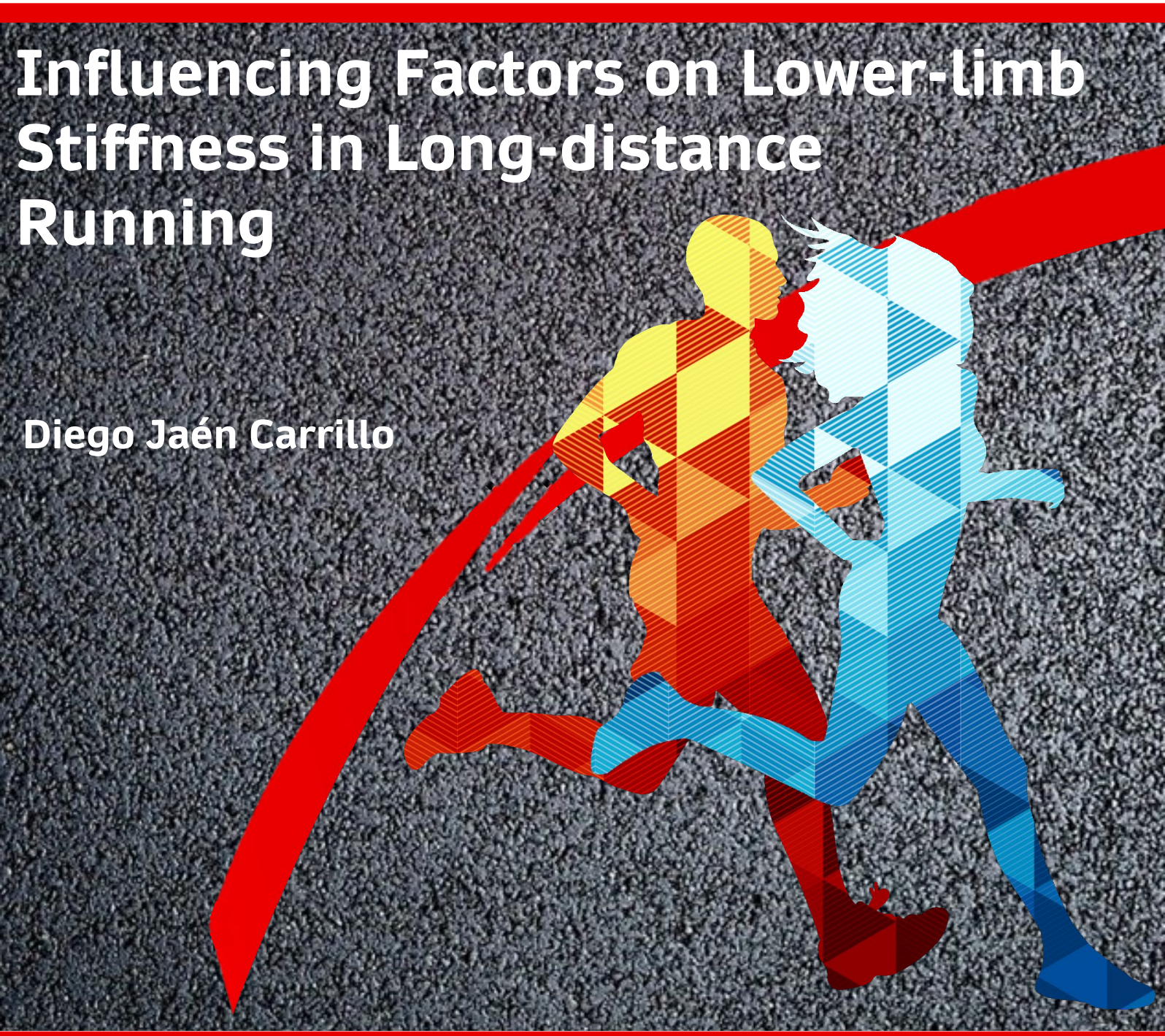
UNIVERSIDAD SAN JORGE
Facultad de Ciencias de la Salud
Programa de Doctorado en Ciencias de la Salud

Influencing Factors on Lower-limb Stiffness in Long-distance Running



Diego Jaén Carrillo

TESIS DOCTORAL

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INFLUENCING FACTORS ON LOWER-LIMB STIFFNESS IN LONG-DISTANCE RUNNING

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"Success is the sum of small efforts,
repeated day in and day out"

Robert Collier

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ABBREVIATIONS

%CT	Percentage of ground contact time over a gait cycle
%FT	Percentage of flight time over a gait cycle
ΔF	Change in force
ΔL	Change in leg length from initial ground contact to mid stance
Δx	Change in length
Δy	COM vertical displacement during ground contact
BMI	Body mass index
BW	Body weight
COM	Centre of mass
CT or tc	Ground contact time
DJ	Drop jump
FFS	Forefoot strike
Fmax	Maximal force
FSP	Foot-strike pattern
FT or t_f	Flight time
g	Acceleration due to gravity
GRF	Ground reaction force
K	Stiffness
Kjoint	Joint stiffness
Kleg	Leg stiffness
Kvert	Vertical stiffness
L	Initial leg length
m	Body mass

MFS	Mid-foot strike
MTU	Muscle-tendon unit
RFS	Rear-foot strike
ROM	Range of motion
RRI	Running-related injuries
RSI	Reactive strength index
SF	Step frequency
SL	Step length
SSC	Stretch-shortening cycle
v	Velocity
vGRF	Vertical ground reaction force
VO2	oxygen consumption
ΔMjoint	Change in joint moment
ΔΘjoint	Change in angular displacement at each joint
Θ	Angle of attack
ω_0	Oscillation frequency

PREFACE

The present PhD Thesis has been organised in 9 different chapters.

Along the **PREFACE** a summary of the subsequent chapters is shown. In chapter **1. INTRODUCTION** running gate and its phases is presented. Moreover, a deep and applied review is presented about the main topic of this PhD Thesis, lower-limb stiffness in running. In chapter **2. HIPOTHESIS** of the current PhD Thesis are described. Chapter **3. AIMS** exposes both general and specific aims. Chapter **4. MATERIAL AND METHODS** exhibits the different methods and protocols used as well as the outcome measures in the different studies addressed to accomplish this PhD Thesis. Chapter **5. RESULTS** is built of a first section where all the studies implemented to accomplish this PhD Thesis are presented in their final version before acceptance or submission and, then, a table is shown to present the main findings of this PhD Thesis. In chapter **6. DISCUSSION** the main findings of this PhD Thesis are critically discussed. **7. LIMITATIONS** describes factors and decisions which might have been limited the outcomes and conclusion of this PhD Thesis. In chapter **8. FUTURE PERSPECTIVE** a proposal of how to research further on this topic is explained. Chapter 9**. CONCLUSION** shows the general conclusions regarding this PhD Thesis. Where needed, **REFERENCES** are shown for each chapter individually. Finally, chapter **10. APPENDIX** contains all the documents related to the research studies.

ABSTRACT

The study of the spring-mass model variables while running resulted in a great contribution to the understanding of the behaviour of such model not only humans, but in animals as well. Although the study of the running spatiotemporal parameters has contributed to obtain a deeper knowledge about the spring-mass model and its capacity to estimate and predict kinematic variables, the contribution of lower-limb stiffness to this model needed further research.

The main aim of the present PhD Thesis was to determine the effect of various influential factors on lower-limb stiffness while treadmill running in healthy adults

Different studies were executed to accomplish the main aim of this PhD Thesis: A unilateral cross-over study aiming at examining the test-retest reliability of the OptoGait photoelectric system for spatiotemporal parameters and lower-body stiffness analysis while treadmill running in healthy adults (Study 1). This first study is key as the entire development of this PhD Thesis has been based on the material and methods implemented and the findings reported; A unilateral cross-over study to clarify the likely relationship between reactive strength index while jumping and lower-limb stiffness while treadmill running in amateur endurance runners as well as sex differences (Study 2); and, ultimately, a unilateral cross-over study to identify the effects of footwear, foot-strike pattern, and step frequency on spatiotemporal parameters and lower-body stiffness (Study 3).

The main findings derived from this PhD Thesis suggest that: the OptoGait system can be used confidently for running spatiotemporal parameters analysis and lower body stiffness at a constant velocity for healthy adults. The spring-mass model reacts differently to tasks based on their specificity principle. Additionally, sex-related differences must be considered when assessing the stretch-shortening cycle. Lower-limb stiffness responds differently to changes in footwear condition, foot-strike pattern, and step frequency.

The findings reported here update the knowledge of lower-body stiffness while running and offer new scopes of action. A reliable and user-friendly system for running spatiotemporal parameters and lower-body stiffness analysis has been provided. Moreover, although both the SSC and lower-limb stiffness are key within the neuromuscular behaviour when elastic energy is used in sport, the specificity principle of each individual sporting task may make them behave differently; additionally, the menstrual cycle should be considered when working with female athletes since musculotendinous properties change over it. Ultimately, it is highly

recommended to avoid measuring the effect of different variables on lower-limb stiffness individually as it has been shown that they influence each another, therefore, the behaviour of the spring-mass model when altering variables such as footwear, FSP, and SF needs to be examined should be analysed attentivel

1. INTRODUCTION

Historically, humans have a need for travelling. Once, this need allowed humans to survive by grouping together in order to harvest, hunt or find a better place to settle down. Humans use to move by either walking or running. The main difference between walking and running, apart from the moving velocity, is the contact pattern between the actor's feet and the ground upon which the action is performed. During walking, there is a double support phase where both feet are in contact with the ground. Running, however, entails a point where neither foot touches the floor, known as the flight phase.

Today, we humans do not run to escape from any predator. Unlike our ancestors, we currently search for active leisure activities where the main requirement is to move precisely to counteract the lack of physical activity derived from the present lifestyle. In the last decades, the simple gesture of tying the running shoes and going for a run is turned into a habit for millions of people of all ages around the world where 43% of respondents in a survey in 2017 claimed that they usually run between 18-40 km per week [1]. Running is becoming more and more popular for several reasons (i.e., pleasure, health or performance) as seen in the increasing number of participants in all levels, from amateur races to the most important marathons in the world, being half-marathon the most attractive event in the U.S. for the 75% respondents of a survey in 2017 [2]. Figure 1 shows the preferred race distant worldwide in 2017 [3].

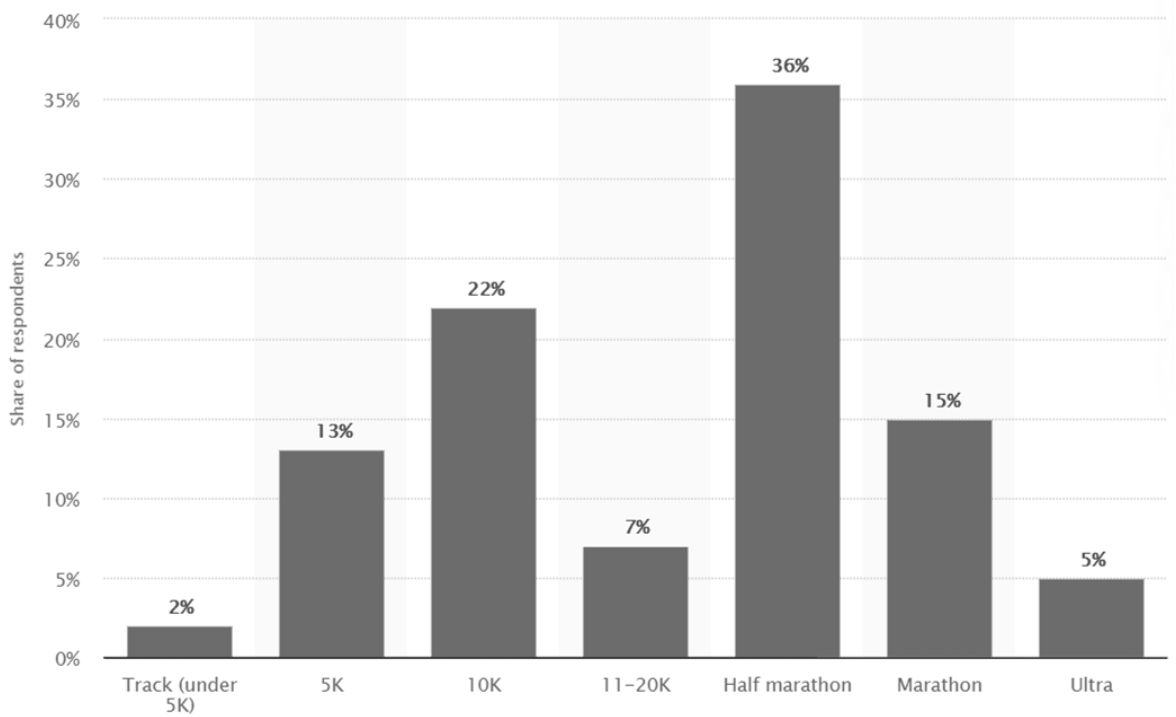

Figure 1. Preferred running race distance worldwide in 2017. Reprinted from Statista GmbH, January 4 2020, retrieved from https://www.statista.com/statistics/933857/running-favorite-race-distance/. Copyright 2020 by Statista GmbH.

Alongside the increasing population participating in running events at all ages and levels has also increased the interest of the scientific community in running research. Consequently, researchers focused on analysing the health benefits derived from running not only physiological, but social and psychological as well. Others intended to clarify the likely mechanisms of running-related injuries (RRI), and many works aimed at determining running performance variables.

1.1. Running gait

Higher velocity, increased step length (SL), a flight phase, and the absence of double stance phase are some of the differences between running and walking [4]. Two airborne phases are present over a running gait cycle resulting in both a reduction of stance phase and increased swing phase. Contrary to walking, the swinging leg and arms created the necessary momentum for forward running [5]. Likewise, running joints and muscles requirements are greater than those for walking (i.e., greater joint range of motion [ROM], particularly for hip flexion, knee flexion, and ankle dorsiflexion) [6].

Mulligan classified running by velocity distinguishing submaximal running (8 – 16 $km \cdot h^{-1}$) and sprinting (velocity > 16 $km \cdot h^{-1}$) [7]. During running, the body's centre of mass (COM) describes a sinusoidal curve in space while the body keeps a forward lean over the entire running gait cycle [4]. Moreover, in order to reduce lateral fluctuations of the COM, the progression line from one step to the next footfall is at or near the centreline [4].

1.1.1. Running gait cycle

The running gait cycle may be classified in different ways. Considering the set of events one single leg describes, the running gait cycle is the span of time from the very early stage of the foot's initial contact with the running service to the moment contact reoccurs. When considering both legs, a running gait cycle is the combination of two consecutive steps, ground contact phase and float phase, respectively. Additionally, three components of the running gait cycle can be defined (i.e., stance phase, swing phase, and float phase) (Figure 2) [6].

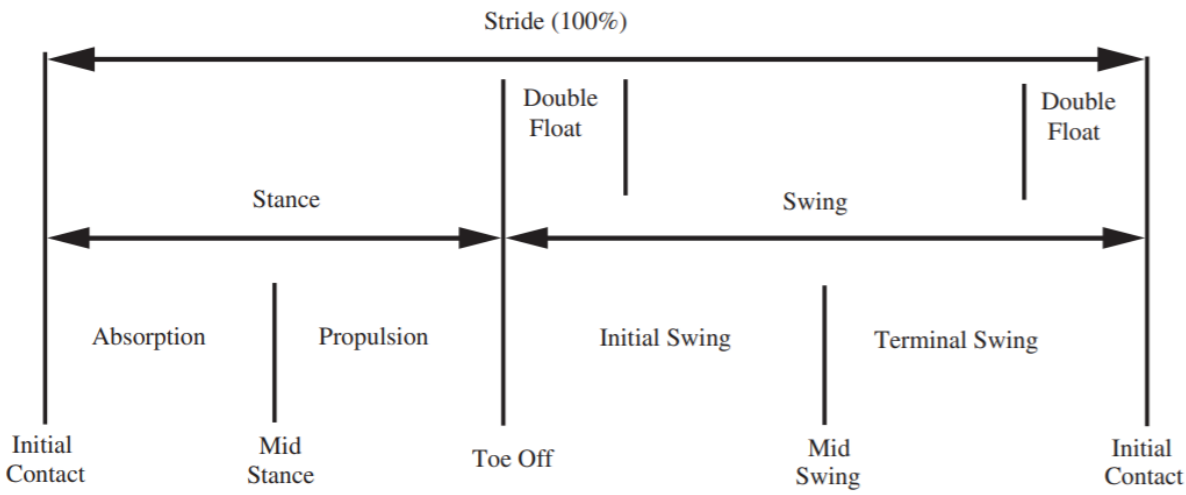

Figure 2. Phases and components of the running gait cycle. Reprinted from 'The biomechanics of walking and running', by S. Ounpuu, 1994, *Clinics in sports medicine*, 13(4), 843-863. Copyright 1994 by Elsevier Inc.

As seen in Figure 2, the stance phase commences with initial force contact absorption and concludes with propulsion. The phase can be deconstructed to three main biomechanical elements: initial contact to mid stance, mid stance to heel-off, and heel-off to toe-off [8]. The swing phase is divided, in turn, into initial and terminal swing preceded and followed, respectively, by a float phase.

Initial contact to mid stance

At this moment the foot collides with the ground. During this phase, energy absorption is the main action of the lower body. The foot-strike pattern (FSP) adopted by a runner contributes to how repeated vertical ground reaction force (vGRF) impacts (~1.5-3 times body weight [BW]) in the first 50 ms of the stance phase are managed [9]. FSPs fall into three main categories: a rear-foot strike (RFS), which sees the heel collide with the ground first; a mid-foot strike (MFS), where the heel and ball land together; and a forefoot strike (FFS) which involves the ball of the foot meeting the ground before the heel [9].

Mid stance to heel-off

There is constant contact between foot and ground throughout this stage. At this moment, the COM reaches its lowest point, marking the end of the absorption subphase of the stance phase. After that, the propulsion subphase happens over the remaining stance phase. The end of this face is discernible as the beginning of supination by raising the heel off the ground.

Heel-off to toe-off

The momentum produced by the swinging leg prepares the stance leg for propulsion. Foot supination begins at heel-off and remains for the rest of the stance phase. Maximum GRF happens when the foot pushes off the ground and propel the body forward [4].

Initial swing

Following toe-off, the body is propelled into the first float phase. Once the float phase is over, the opposite foot strikes the ground and the mechanism described above is repeated.

Terminal swing

Here, the swinging leg is about to contact the ground along the progression line. At initial contact, a complete running gait cycle occurs, and the patterns above described start again when the following cycle begins.

1.2. Spring-mass model

During the running stance phase, the leg function resembles the behaviour of a spring which compresses and decompresses continually [10], being the body mass (m) responsible for such leg-spring compression [11] (Figure 3). This single linear leg spring and a mass equivalent to that of the runner has been shown to describe and predict the mechanics of running extraordinary well [10,12-15]. Mechanical energy is stored over the leg-spring compression, represented by the eccentric phase of stance, whereas the concentric phase of stance releases the stored energy as elastic energy [11].

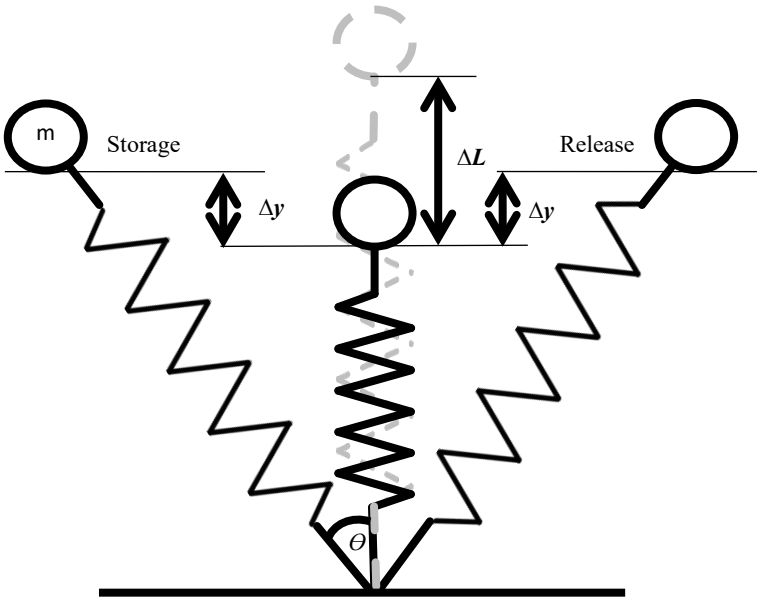

Figure 3. Diagram of the spring-mass model with a mass fixed to a mass-less spring representing initial contact, mid stance (COM at its lowest point), and propulsion phases while running. Adapted from 'Leg power and hopping stiffness: relationship with sprint running performance', by S.M. Chelly and C. Denis, 2001. *Medicine & Science in Sports & Exercise*, 33(2), 326-333. Copyright 2001 by Lippincott Williams

This model consists of a mass representing the COM of the runner attached to a mass-less spring. Contact phases in which the system rotates forward over a monopodial support alternate with float phases where the system behaves ballistically [16]. Initial angle of attack (Θ) and the stiffness of the leg-spring play key roles in the final behaviour of such spring-mass model.

The stretch-shortening cycle (SSC) [17] and lower-limb stiffness [18], particularly vertical stiffness (Kvert) and leg stiffness (Kleg), are the two most important neuromuscular elements linked to elastic energy use. Therefore, the influence of lower-limb stiffness in running using the spring-mass approach is deeply analysed along this PhD Thesis. While pursuing this PhD Thesis, a narrative review has been developed to identify influencing factors on lower-limb stiffness during horizontal running and to discuss these factors from both an injury prevention and performance perspective, respectively (**Appendix 2**).

1.3. Stiffness from the sine-wave model

The term stiffness originated in physics, as part of Hooke's Law. Hooke's Law is defined as $F=kx$, where F is the force required to deform an object, k is the proportionality constant, and x is the distance the object is deformed [19]. The proportionality constant, k, is also known as the spring constant, and it describes the stiffness of an ideal spring mass system [19]. This is derived by rearranging the aforementioned formula to $k = \Delta F/\Delta x$, where ΔF is change in force and Δx is change in length [20]. In other words, when an object that obeys Hooke's Law deforms, the change in length will be directly proportional to the force acting upon said object [21]. Reducing it to its simplest definition, stiffness describes the relationship between a given force and the magnitude of deformation of an object or body [19,22].

Stiffness is a concept frequently used to characterise human movement or describe neuromuscular function [19,20,23,24]. With regard to the human body, stiffness can be described at a broad range of levels, from that of a single muscle fibre to modelling the entire body as spring-mass system [19,22]. The main responsible for the leg-spring compression is the body's mass [25]. Specifically, the eccentric phase of stance represents the compression of the leg-spring, during which mechanical energy is being stored [25]. The concentric phase of stance represents the decompression of the leg-spring, which is accompanied by the release of the stored mechanical energy in the form of elastic energy [25]. Given its elastic properties, the leg-spring, as every conventional spring, tends to resist to any deforming force. The magnitude of this resistance depends on the leg-spring's stiffness. Similarly, at the leg-spring decompression the magnitude of the returned elastic energy is positively correlates with stiffness [26]. Greater

stiffness of the musculotendinous unit would be anticipated to maximise the conversion of potential energy, stored within the elastic components of the lower limb during eccentric lengthening, to kinetic energy released during subsequent contractile shortening [27]. Indeed, a simple spring-mass system has been successfully used to model the basic mechanics of a variety of multi-jointed whole-body movements that involve the SSC (e.g., running, jumping and hopping) [28]. The spring-mass model comprises a point mass (equal to body mass), which is supported by a single Hookean spring –representing the leg [14,15].

When exploring the relationship between stiffness and athletic performance, four measurements are commonly utilised [28,29]:

1. *Vertical stiffness* (kN/m) describes the global compression of a runner [30], that is, the vertical displacement of the centre of mass in response to vertical GRF during a task performed in the sagittal plane [20].

2. *Leg stiffness* (kN/m) describes the mechanical behaviour of the leg spring (i.e., muscles, tendons, and ligaments) as it is compressed during the early phase of stance [15,30]. From a functional perspective, Kleg is the summative lower-limb stiffness that influences performance during multi-jointed whole-body movements involving the SSC.

3. *Joint stiffness* (Kjoint, Nm/rad) describes the angular displacement of a joint in response to the moment at the joint [31].

4. *Musculotendinous stiffness* is calculated using the oscillation technique; an active and loaded muscle-tendon unit is perturbed and its free response is recorded [28].

As mentioned above, stiffness is a product of both force and length, therefore Kvert and Kleg are calculated as the ratio of the peak vGRF to peak COM displacement and peak leg compression respectively during the period of ground contact [15]. It is worth noting, however, that Kvert and Kleg are identical when the COM moves purely in the vertical direction [15,19]. Yet, it has been proposed that Kvert is not able to detect the different contribution level of each joint to the determination of the whole leg's stiffness [32]. For this reason, the concept of Kjoint was introduced, which models the relationship between joint moment and joint angle [25]. Hence stiffness can be measured either for the whole-body system or for each joint in the system, but can also be measured passively, i.e. when muscles are not producing force. It has been argued that passive stiffness has less to offer the research community in that most lower body sports

involve muscles producing force, and it is muscle force which contributes most significantly to lower body [33]. However, its contribution might be equally interesting.

As Kvert and Kleg influence the regulation of both spatiotemporal and kinematic variables, they are usually used while identifying these characteristics in individual runners. During SSC movements, Kjoint has been shown to be the primary determinant of Kleg [34-39]. For the calculation of Kjoint, the vertical load in the case of Kvert is substituted by the joint moment value, and the vertical displacement of the COM is replaced by the change in the joint angle under the general concept of a torsion spring [40]. Joint angles at touchdown have also been shown to influence Kleg during SSC movements [35,37,41-43], as greater alignment of the vertical GRF vector relative to the joints reduces the moment arm of the GRF, thus increasing Kleg [44].

The reactive strength index (RSI) has been commonly utilised as a means to quantify either SSC or plyometric performance in the practical strength and conditioning setting [45,46]. The RSI is defined as the ability to switch rapidly from an eccentric to a concentric contraction [46]. Since the RSI can be measured from the ratio of flight time (FT) and contact time (CT), both variables play a determinant role within the RSI score. It has been proved that Kvert correlates highly with reactive strength index (RSI) while jumping [47]. The RSI is a measure that is often used to quantify dynamic lower extremity during a drop jump (DJ) [46,48,49] and exhibits a highly reliable (i.e., ICC > 0.90) and simple index of performance that is also easy to measure and interpret [50-52]. As RSI is measured, the instructions that are provided to athletes during DJ and RSI testing are typically "Jump as high and as fast as you can" [48]. These instructions are given so as to encourage athletes to maximize jump height and minimize ground contact time, which in combination optimize RSI. Considering that these instructions likely lead to large GRFs over small periods of time, it could be hypothesized that the RSI is associated with vertical stiffness [53,54]. Kipp and colleagues (2017) analysed the correlations between RSI and biomechanical variables, focused on athlete's Kvert and COM, during the eccentric and concentric phases of the DJ [47]. Vertical stiffness during DJ were significantly correlated to RSI at each of the three drop heights (30, 45, and 60 cm) [47]. Since the RSI in the aforementioned study was strongly associated with vertical stiffness at each drop height, and did not change across heights [47], it appears to be closely linked to the vertical stiffness of the body's musculoskeletal system during DJ.

Despite several methods for lower-limb stiffness analysis have been proposed [55-60], Morin's sine-wave model (51) has been widely used for Kvert and Kleg determination due to

its accuracy, efficacy, and the reduced amount of information it requires gathering (speed, leg length, CT, FT, and m). According to Morin's method, Kleg is estimated as follows

$$K = \frac{F_{max}}{\Delta L} \text{ (Equation 8)}$$

ΔL represents how leg length changes from primary ground contact to mid stance, where initial leg length (*L*) was measured as the distance in metres between the greater trochanter and the ground while standing upright and shod. *ΔL* was derived using the equation

$$\Delta L = L - \sqrt{L^2 - \left(\frac{vt_c}{2}\right)^2} + \Delta y \text{ (Equation 9)}$$

Δy signifies the vertical displacement in metres of the runner's COM during ground contact; t_c represents duration of ground contact in seconds; and *v* represents sept averaged running velocity ($m \cdot s^{-1}$). The following equation determined Δy_c:

$$\Delta y_c = \frac{F_{max} {t_c}^2}{m\pi^2} + g\ \frac{{t_c}^2}{8} \text{ (Equation 10)}$$

where *m* denotes the participant's body mass in kilograms and *g* is the acceleration due to gravity (-9.81 m s^{-2}). The modelled peak vertical force during contact was found as follows:

$$F_{max} = mg\ \frac{\pi}{2}\left(\frac{t_f}{t_c} + 1\right) \text{ (Equation 11)}$$

with t_f representing flight time.

As mentioned above, Morin's sine-wave model has been broadly used for both Kvert and Kleg analysis since it has reported accurate lower-limb stiffness values. However, its test-retest reliability for treadmill running is still unknown, thus, a research study was designed and implemented to determine it (**Study 1**).

1.4. Measurements of stiffness

1.4.1. Vertical stiffness

Vertical stiffness is typically considered the quotient of maximum ground reaction force (GRF) and COM displacement [34,44,56,57,61-69]. That is:

$$K_{vert} = \frac{F_{max}}{\Delta_y} \text{ (Equation 1)}$$

In most studies maximum GRF was measured using force platforms [34,44,56,57,61-69], and COM displacement was obtained by double integration of vertical acceleration as described by McMahon and Cheng [15,34,64,70], or by Cavagna [56,57,61-63,65,69,71,72]. Two papers which measured GRF using a force platform were unclear about the method by which calculation of COM displacement was determined [44,73].

Numerous studies used an identical approach to measuring Kvert using force plate, pressure sensor or accelerometer technology, but modelled COM displacement [37,57,68] via independent variables including ground contact time, flight time etc. The relevant calculation is expressed as:

$$t_f = \left(\frac{t_c + T_f}{2}\right) - t_c \text{ (Equation 2)}$$

$$\Delta_y = \frac{F_{max} t_c^2}{m[\pi^2]} + g\frac{t_c^2}{8} \text{ (Equation 3)}$$

Where accelerometer technology was employed, GRF was also modelled [67,74]:

$$F_{max} = mg \cdot \frac{\pi}{2} \cdot \left(\frac{t_f}{t_c} + 1\right) \text{ (Equation 4)}$$

Another model of Kvert described [43] was:

$$K_{vert} = m\omega_0^2 \text{ (Equation 5)}$$

Resolving the natural frequency of a mass spring system representing a body using:

$$\frac{F}{mg} = \left(\frac{u\omega_0}{g}\right) sin\omega_0 + 1 - \cos\omega_0 t \text{ (Equation 6)}$$

The most common model of Kvert used was the quotient of maximum GRF and COM displacement (Equation 1). The proposed alternative above (Equation 5) was not adopted for use in later research and did not present any descriptive statistics [43]. Therefore, a qualitative analysis regarding the accuracy of the model could not be determined.

For studies which incorporated GRF and COM displacement (Equation 1) the main difference in measurement methodology was in calculating the latter; some used a method described by McMahon and Cheng [15], others used a method described by Cavagna [72]. Standard deviation is the only statistical value which it is possible to compare across the methods of all the studies. For studies which measured COM displacement using McMahon and Cheng's method [15], the standard deviation of Kvert as a proportion of the mean was similar to results obtained using Cavagna´s method [72]. Absolute Kvert and standard deviation from the Morin et al. study [57], which used Cavagna's method [72] to measure COM displacement, are reasonably similar to those recorded by Hunter and Smith's study [70], which used McMahon and Cheng's method [15]. Each study required participants to execute similar tasks. It is of note, however, that where Cavagna's method was used, participants were asked to engage in a hopping or running, whereas the hopping task was absent from McMahon and Cheng's method.

Kvert was typically greater when performing activities which required greater force production (e.g. faster running velocities, single leg hopping as opposed to a double leg); however, more measurement variation was also recorded. The implication here is that reliability issues may arise, or larger sample sizes might be required. Finally, similar results were produced by studies measuring Kvert via the model of the quotient of GRF and COM displacement (Equation 1) which also modelled GRF and/or COM displacement, and those where GRF and COM were directly measured. Hence modelling the stated variables for measuring Kvert may provide a suitable 'option' where direct measurement limitations exist, as in the Morin and colleagues study which revealed a small bias for results when GRF and COM displacement was modelled as opposed to measured [57].

1.4.2. Leg stiffness

Comparison can be drawn between the lower extremity while running and the stiffness characteristics of a spring [14,15]. Kleg is frequently determined as the relationship between peak vertical ground reaction divided by relative compression of the leg during ground contact [19]. Kleg may serve as a global surrogate for loading rate and the subsequent kinematic response of the lower extremity during running. In this substitution, lower Kleg is associated with an increase in both joint excursion and reliance upon active muscle contributions to modulate landing tasks [75]. Higher Kleg is associated with reduced joint excursion and increased impulsive loading to bones and cartilage [76,77]. A link has also been proposed between heightened Kleg and increased lower extremity injury occurrence [78]. Indeed, Pruyn and colleagues [79] found that vaster variations in Kleg between legs in Australian rules football

players was also prospectively associated with more incidents of lower extremity injury. Kleg may therefore be influential in injury rates during dynamic activities.

Kleg is multifactorial in nature, engaging numerous active and passive characteristics of the musculoskeletal system [20,31]. It is traditionally assessed utilizing motion capture and inverse dynamics, but this is not clinically feasible due to the training, time and cost demands involved [19]. It is possible to estimate clinically during hopping but still requires specific equipment and training [80-82]. Previously, the stiffness of the hip, knee and ankle joints has been shown to influence Kleg, including contributions from both passive and active structures [83].

1.4.3. Joint stiffness

Overall, joint stiffness is responsible for Kleg. Kjoint, defined as the ratio of the maximal joint moment to the maximum joint flexion at the middle of the stance phase [19], can be calculated with the torsional spring model [36]. The model assumes four rigid segments (foot, shank, thigh and head-arm-trunk) interconnect with torsional springs of the hip, knee and ankle. Joints of the lower extremity flex across the period which is initiated by the instance of touch down until the middle of the ground contact phase. Therefore, Kjoint of the hip, knee and ankle is calculated as

$$K_{joint} = \frac{\Delta M_{joint}}{\Delta \theta_{joint}} \text{ (Equation 7)}$$

where ΔM_{joint} and $\Delta \Theta_{joint}$ signify changes in joint moment, and the angular displacement at each joint during the first half of the stance, respectively [36]. There are several studies that have calculated Kjoint during human running [34,39,84-86]. Four of them kinematically analysed Kjoint using high-speed video cameras and force plates to find horizontal and vertical forces (29, 36, 42, 63). Arampatzis and colleagues developed their own method of Kjoint calculation during human running [34]. The ratio of negative mechanical work to the change in joint angle established Kjoint, and both kinetic and kinematic analyses determined work and the change in joint angle. The method has been questioned by Günther and Blickhan [85], who posed that it was unreasonable to divide a work integral by a change in joint angle in order to calculate stiffness. Four of the aforementioned studies [34,39,84,85] concluded that as running speed rose, ankle Kjoint remained constant and knee Kjoint increased, giving rise to the conclusion that knee Kjoint is the major modulator of Kleg during running [34,39,84,85].

Nevertheless, it is the ankle which plays a dominant role in storing and generating propulsion energy during the stance phase of running [87,88]. The ankle shows first a loading state,

during which the internal plantarflexor moment rises during dorsiflexion, and the periarticular joint structures absorb energy [89]. An unloading state follows in which the plantarflexion moment decreases while the joint plantarflexes, and the periarticular joint structures produce energy [89]. The level of stiffness (i.e., the variation of joint moment per unit of joint angle variation) can be affected both by structural adaptations of the muscle–tendon units surrounding the joint and by neural adaptations which control instantly the characteristics of these muscle–tendon units [90-92]. For instance, habitual forefoot strikers who usually land with a plantar-flexed ankle long-term demonstrated adaptations in muscle and tendon architecture in the lower limb. Such adaptations included shorter gastrocnemius medialis fascicles [93], thicker Achilles tendon [94] and stiffer foot arch [95] and could result in different load distribution in the muscle–tendon unit [96]. That is, the role of the elastic components is elevated and muscle fibres contraction slows, which is advantageous for maximal power output and efficiency [97].

1.4.4. Musculotendinous stiffness

In human movements, musculotendinous structure as a functional unit (i.e., muscle-tendon unit [MTU]) of the lower limbs usually shortens immediately subsequent to lengthening, and this extends to running [98]. The alternating action of lengthening and shortening describes the so-called SSC. It has been established that the storage and subsequent release of elastic energy in the SSC enhance the mechanical efficiency and power output of the MTU [99], and thus SSC performance is susceptible to MTU elasticity [98]. The MTU elasticity in SSC exercises is suggested to be reliant upon neuromuscular factors and intrinsic MTU stiffness [100]. Owing to the tendon's lack of rigidity, elastic energy can be stored when the MTU is lengthened, thereby enhancing MTU performance in SSC exercises [99,101,102]. Regarding stiffness, focus has been placed upon the functional roles of the MTU and tendon stiffness of knee extensors and plantar flexors in sprinting and running [37,103,104].

MTU stiffness can be assessed by applying the oscillation technique [54,105-107], which sees an active and loaded MTU perturbed and the free response recorded; thus, the human muscle was modelled as a damped spring system. Systematic perturbation will result in damped oscillations [28].

1.4.5. Passive stiffness

If running performance correlates with passive stiffness, measuring the latter could be useful in assessing running performance and evaluating the effects of training and/or rehabilitation [108]. Evaluating simple passive stiffness is relatively easy and can be achieved

with the use of an isokinetic dynamometer [108-110]. For this calculation, the chosen joint of the subject, who is placed and secured in a dynamometer, is taken through a range of motion which is not actively resisted. This measured passive resistance torque, the ratio of which is calculated against angular displacement as a measure of stiffness [109,110] at hip, knee, and ankle joints.

Spurrs and colleagues [111] previously reported correspondence between improved running economy by 6 weeks of plyometric training and enhanced passive stiffness in plantar flexors. The mechanical properties of the MTU has been frequently investigated using passive stiffness, even though as a parameter it encompasses the properties of other tissues including skin, subcutaneous fat, fascia, ligament, joint capsule, and cartilage [112]. Kubo et al. [113] suggested that passive stiffness in the plantar flexors may in fact be reflective of muscle, rather than tendon, tissue properties. Recent studies demonstrate that passive plantar flexor stiffness and muscle stiffness measured using shear wave elastography [114,115] are correlational. These findings indicate that passive stiffness bears more of a relationship with the properties of muscle tissue than tendon tissue. Ergo, passive stiffness may be useful for evaluating muscular flexibility [108]. Ueno and colleagues [108] reported a connection between greater passive stiffness of the plantar flexors and better endurance running performance, demonstrating significantly higher passive plantar flexor stiffness in well-trained endurance runners than in their untrained counterparts [108]. Increased passive plantar flexor stiffness was also found among faster runners as opposed to in the slower group [108].

1.5. Stiffness and Running performance

The ability to instigate heightened lower limb stiffness is likely most beneficial to activities where transmitting a given impulse in a shorter time period would be advantageous, such as maximum velocity running [116]. Although it may seem reasonable to assume a relationship between lower-limb stiffness and athletic performance exists, the evidence base is perhaps not as definitive as coaches and practitioners tend to perceive. Indeed, relevant literaure is divided and inconsistent on the matter. Attempts have previously been made to outline the different measurements and methods by which lower-limb stiffness can be calculated [23,28].

One widely used model to estimate the stiffness of the body during human movement is that of a simple spring-mass [14,15,19,23,53,117-120]. In this model, as described above, the lower limb is embodied as a simple 'leg-spring' supporting the mass of the body [14,19]. Stiffness in tasks such as hopping [119] and running gait [117]. has been calculated via this method. The spring-mass model assumes a linear relationship between COM displacement and GRF, therefore peak

displacement and force should coincide [19]. The application of the spring-mass model to predicting a given activity is evaluable through calculation of the correlation coefficient between force and displacement, thus rendering these criteria relevant to individual trials. Investigations into opping, a task to which the prospective relevance of the model will be discussedin subsequent sections, have entailed conservative inclusion criteria ($r \geq 0.8$). However, to model sprinting gait and deviation from the spring-mass model, the higher value ($r^2 \geq 0.9$) has been proposed [121].

Both Kvert and Kleg calculations have been reported during gait-based investigations, though the two measurements may yield disparate relationships. Proportionate increase has been recorded between Kvert and running velocity [39,57,65,68,122] and stride frequency [56]. However, whilst Arampatzis and colleagues claimed to have witnessed increases in both vertical and Kleg with running velocity [34] , still others demonstrated that Kleg is unaffected by running velocity [57,65,122]. incongruent results suggest that Kvert measurement is potentially more sensitive than than Kleg in studies aiming to explore relationships with running performance. The position is also supported by the findings of further studies, such as one which reviewed the impact of fatigue and concluded that reductions in repeated sprint velocity were mirrored by reductions in Kvert, however, Kleg was unchanged [68]. Girard and colleagues garnered similar results from 800-m track running [123]. Nagahara and Zushi also reportedly observed training-induced improvements in Kvert and performance in sprinters, but static Kleg [124]. Yet the inverse may be true for slower velocity, longer duration running; several studies have reported lowered Kleg and minimal change in Kvert following fatiguing protocols [125-128].

Morin and colleagues established that fluctuations in the time spent in contact with the ground, which was manipulated in the study, explained a larger proportion of variance in changes in Kleg than changes in stride frequency ($r^2 = 0.90$ and 0.47, respectively) [129]. Though the metabolic cost of running was not under consideration, greater Kleg has been seen to be correlational with lower metabolic cost [26] and is therefore seen as an economical running strategy [18]. Consequently, it is arguable that producing increased Kleg and concurrently maintaining stride frequency, facilitated by shorter CT, would limit the metabolic cost of running. Aside from the aforementioned Morin's study, there is a lack of focus on CT and Kleg from within-participant study designs investigating economical running.

A recent study [130] identified that ground contact time and Kleg as self-optimised gait characteristics of running, revealing that trained runners perform at, or approach, their

mathematical economical optimal during submaximal running. Furthermore, 90% of the runners used self-selected CT and Kleg, exhibiting metabolic costs within 5% of their optimum [130].

When comparing sprinting and endurance athletes while hopping [131] and 20-m progressive run and 30-m sprint [132], three factors were elevated in sprinters: Kleg during hopping at 1.5 and 3.0 Hz and DJs from 30 cm; knee stiffness during hopping at 1.5 Hz; and ankle stiffness during hopping at 3.0 Hz. Sprinters additionally possessed shorter CT and longer FT at both hopping frequencies studied [131], which is indicative of greater reactive strength capacity. The implication herein is that sprinters may run with a greater stride length for a given CT than endurance athletes and the former can therefore achieve much greater running speeds than the latter. The aforementioned differences in Kleg and Kjoint between the two athletic groups have been ascribed to greater Achilles tendon stiffness observed in sprinters [133]. However, as the sprinters may have consistently performed more strength and power training, it is plausible that increased relative strength capacity and greater SSC utilization may also be contributing factors to the greater Kleg reported for this group.

Despite their comparatively lower Kleg, endurance trained athletes nevertheless showed greater Kleg than untrained subjects during hopping at 2.2 Hz [86]. However, unlike the proposed explanation for the greater Kleg in sprinters, any differences in Achilles and patellar tendon stiffness between endurance athletes and untrained subjects have been refuted by the literature [133-136]. A possible explanation for greater Kleg in the endurance group, therefore, may be explicable as the greater prevalence of slow-twitch muscle fibres which results from endurance training [137]. For example, when muscle fibres were compared, slow-twitch demonstrated greater dynamic stiffness than fast-twitch [138]. What is more, endurance training induced increased in muscle stiffness, which was associated with a decrease in fast-twitch muscle fibres [139].

A stiffer leg would potentially store and release energy more effectively, which could subsequently reduce the metabolic cost of running [130]. However, increases in both Kvert and Kleg have been linked to increased task intensity and improved task performance [29]. Kvert sensitivity may be more responsive to change than Kleg in high-velocity tasks, whilst in exhaustive running Kleg may be more susceptible [29]. As the likelihood of an existing relationship between RSI and lower-limb stiffness within endurance runners is still unclear, a unilateral crossover study was executed to clarify the behaviour of both variables (**Study 2**).

1.6. Stiffness and Running-related injuries

Running-related injuries are multifactorial. Repeatedly applying high-impact forces without sufficient intervals may lead to injury [140,141]. High forces exerted on the lower extremity tissues during running [140,142,143], alongside behavioural (eg, training history, injury history) [144-148] and physiologic risk factors (i.e., quadriceps and hamstring flexibility, quadriceps angle , arch height, and strengths of muscle, bone, and other tissues) [149,150] constitute potential etiologic factors. While this general model of overuse injury has attracted some support [141], previous retrospective works showed that free-injury runners had greater GRFs than injured runners [147,149-151]. A few research groups employed an alternate design (i.e., prospective longitudinal), enhanced by observing initially uninjured runners to evade the limitations of previous studies. At 2-year follow-up, Davis and colleagues [142] were able to specify that there was no main effect of joint loading between female runners who sustained an overuse running injury versus those who were without injury. Buist and colleagues [152] identified a connection in male runners between etiologic factors (higher body mass index [BMI], previous injury, previous sports participation without axial loading) and injury as compared with female runners (navicular drop) after 13 weeks of training. Nielsen and colleagues [153] conducted a 1-year observation period after which no difference in rear-foot motion between injured and injury-free runners manifested.

Messier and colleagues determined the risk factors for runners, distinguishing between uninjured amateurs and those diagnosed with overuse running injury runners, across a 2-year observational period [151]. They found similar distribution of observed injuries as indicated by prior studies [142,148,154] with the knee and foot the most common injury sites [151]. Macera and colleagues [145] little gender distinction in overuse running injury rates, recording approximately 50% per year for men and women, whereas Taunton and colleagues [148] noted that the frequency of some injuries was sex dependent. Meanwhile, Messier's data indicated that gender was significant in predicting injury incidence: female runners were injured more often than male runners (73% vs 62%, p = .046), and approximately half of each sustained injury more than once during the 2-year period [151]. Conversely, a cohort study of 532 novice runners training for 13 weeks in preparation for a 4-mile race saw 1.5 times more men than women injured (hazard ratio = 1.5, P = .04) and had different etiologic factors [152].

Previous work reported that the lone noteworthy predictor of injury in their multivariable analysis was maximum knee stiffness, being significantly higher in the injured group after controlling for training pace and body mass [151]. In fact, both knee stiffness and body

mass highly correlated within the injured group [151]. It has been suggested that greater knee stiffness, which is more prevalent in runners with higher BW (≥80 kg), significantly raises the possibility of suffering an overuse running injury [151]. As knee stiffness involves aspects of motion (knee flexion angle) and force (knee extensor moment), a viable means of predicting overuse running injuries would, unsurprisingly, incorporate both factors.

1.7. Influencing factors

Many are the factors that may influence lower-limb stiffness while running. The way a runner's foot collides the ground, either the presence or absence of footwear, or the type of surface where one runs are just a few examples of stiffness influential factors. Along this section, the most influential elements affecting lower-limb stiffness during running are described as well as their relationship with Kvert and Kleg. In literature, others have determined the influence of these factors individually on lower-limb stiffness without considering that each of these variables influences the others and, therefore, lower-limb stiffness. Given that, another unilateral crossover study was implemented to define the influence of the footwear condition, foot-strike pattern (FSP), and step frequency on the running spatiotemporal parameters and both Kvert and Kleg during treadmill running (**Study 3**).

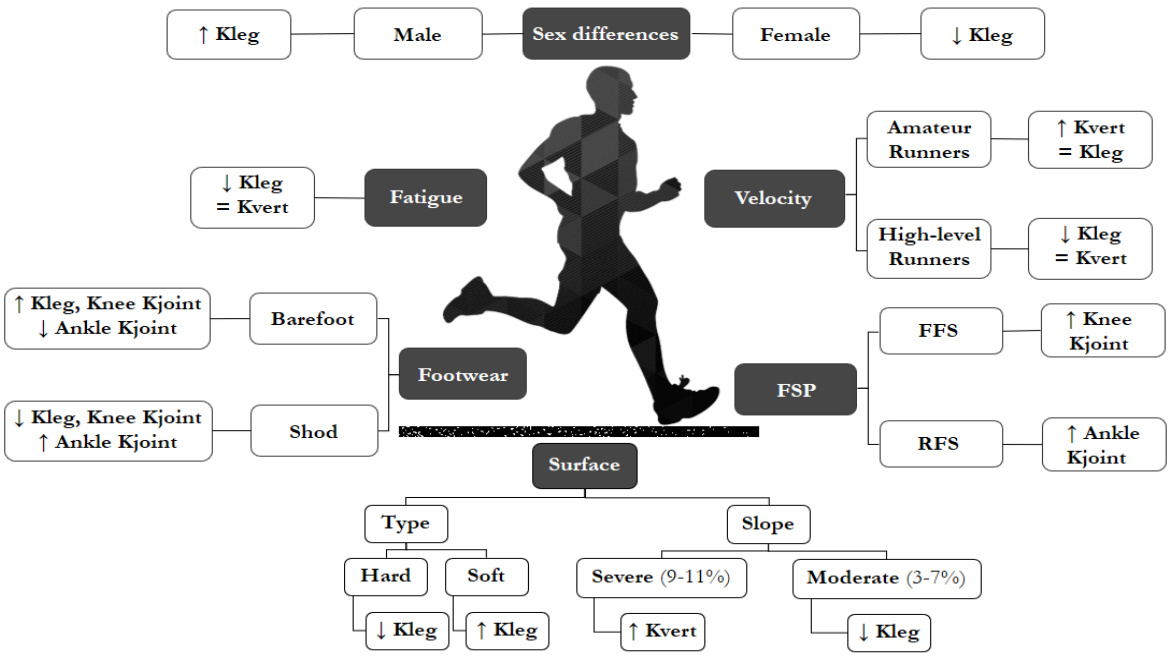

Figure 4. Diagram for the influencing factors on lower-limb stiffness while running.

1.7.1. Foot-strike pattern

The FSP seems to be related with behaviour of the different lower-limb stiffness assessments [9,32,155-157]. Particularly influential is the forefoot striking pattern, in which the ball of the foot meets the ground ahead of the heel [9]. This FSP has been linked to effects on the knee, specifically increased Kjoint and decreased range of motion (ROM). Focusing on the ankle, the pattern results in a decrease of Kjoint and increased ROM [25]. Regarding a rear-foot striking (RFS) pattern, the relationships are reversed [9,32,155,156]. As ratio of joint moment to angle ($\Delta M/\Delta\theta$) regulate these Kjoint measurements, the observations are potentially mechanistically explicable as follows: in terms of forefoot striking (FFS), the increased ROM is responsible for the decreased Kjoint at the ankle, while the reduced ROM is responsible for the elevated Kjoint at the knee [25]. As before, the inverse is true in the case of a RFS [156]. These interactions, as detailed above, describe the effect of the striking pattern on Kjoint measurements. From a wider perspective, Kleg is a better descriptor of the spring-like behaviour of the leg during running than the individual stiffness of each joint [25]. Hence it would prove beneficial to examine the effect on the magnitude of global Kleg ankle and knee Kjoint value variation as they interact with each striking pattern.

The relevant literature appears to be divided [36,77,155]. Farley and Morgenroth postulated that Kleg is more sensitive to ankle Kjoint [36] , a selective sensitivity apparently due to the leg's geometry. Horizontally oriented foot length creates a longer GRF moment arm which is related with higher moment and angular displacements relative to the knee and hip joints [36]. Thus, an absolute increase in ankle Kjoint by a RFS pattern would hold more influence over global Kleg [25]. However, the experimental intervention of this study was a hopping task, which intrinsically involves a FFS pattern. On the other hand, others imputed the knee as the joint of greater influence on Kleg [77,155]. Contrary to Farley and Morgenroth's study where participants used a FFS pattern [36], in Williams and colleagues study all the runners exhibit a RFS pattern [77]. The simultaneous increase in knee Kjoint and Kleg, in addition to the observed ankle Kjoint reduction during the forefoot running experimental condition of the study, was used as an argument [155]. Hamill and colleagues [32] were seemingly the only researchers evaluating change in ankle Kjoint in two groups of runners with distinct foot strike patterns. Based on two factors, precisely the presence of an impact peak on the vertical GRF the ankle angle at landing, participants were classified as either RFS or FFS runners. Runners may have been misclassified according to the selected criteria [158]. Nevertheless, Hamill and Gruber [32] reported more compliant ankles and more (negative) work absorption in forefoot strikers compared to habitual

rearfoot strikers when running with their preferred FSP (FFS). However, no differences were found with habitual rearfoot strikers running with a FFS pattern (non-preferred mode).

Most studies with the purpose of examining FSP impact on running economy suggest that striking pattern, and the previously discussed effect on stiffness measures, do not influence running economy [159-162]. Gruber and colleagues determined from their results that adopting a RFS pattern has potential advantages for running at faster speeds [161]. However, the recorded percentage differences favouring the RFS pattern were not considered sufficient to introduce physiologically beneficial running economy modifications for the average recreational runner. Yet a RFS pattern was suggested to be favourable for elite athletes, given that any running economy improvement may be of determining importance in terms of competition performance [161]. A corresponding modelling study posed that a RFS pattern was optimal in shod conditions in terms of energy conservation, while non-RFS patterns were preferable for equally sharing muscle work and minimizing local muscle fatigue [163]. According to the study, RFS running was characterized as the most "multipurpose" running pattern and this adaptability was used as a possible justification for its predominance across runners of different levels and backgrounds [163] , with 75% of runners making initial ground contact heel first [161]. Thus, the mechanical advantages of FFS are counterbalanced. Although the pattern appears to enhance the leg's ability to store and reutilize elastic energy, heightened muscle activation requirements lead to increased contractile costs, especially in the triceps surge muscle group, and no differences can be observed between FFS and RFS patterns [162].

1.7.2. Footwear

Another apparently prominent feature in regulating lower-limb stiffness measures is running shoes. Some of the surveyed runners replaced their previous heel-cushioning models with minimalist shoes [164] , a variable with implications for several biomechanical aspects. Firstly, as running barefoot tends to strike mid-foot or forefoot, stride length is modified, consequently influencing loading rate, plantar peak pressure, step frequency (SF), muscular activity, leg compliance, ankle, knee and hip kinematics [9,61,165-167]. Despite this general inclination towards flatter foot placement on landing when transitioning from shod to barefoot running, there remain barefoot runners with heel-to-toe contact pattern [168].

Regarding the relationship between footwear and lower extremity stiffness, studies have compared differences in: Kleg when running barefoot versus traditional shoes [169]; leg [170] or joint [171] stiffness with varying midsole hardness in traditional running shoes; Kleg [172] or Kjoint

[173] when running in minimalist compared to traditional running shoes. Additionally, recent findings revealed that after a 4-week adaptation period, runners wearing fully minimalist shoes demonstrate higher vertical and Kleg than runners wearing ultra-cushioning shoes [174]. Cumulatively, these studies clearly show that footwear influences Kvert, Kleg, and Kjoint. Additionally, Jing and colleagues reported that while shod Kvert, Kleg, and knee stiffness decrease, both hip and ankle stiffness increase in comparison with barefoot [175]. Emphasis of existing literature has largely been on effect of footwear condition on both Kleg and Kvert. The most salient finding of all the reviewed studies was that barefoot running or running at minimalist footwear is accompanied by increases in Kleg [61,157,169,172,176]. Given that Kleg is expressed by the ratio of vGRFs to the magnitude of the leg compression expressed as ΔL [172], the Kleg increases recorded in barefoot and in minimally-shod running conditions are attributable either to decreased leg compression through shorter CT or to increased vGRF values [25].

Another approach centres on the interplay between Kleg and Kvert. Shih and collegues (2013) and Lussiana and colleagues (2015) reported that running barefoot or in minimal shoes was followed by an increase in Kleg, with not significan Kvert variation [157,172]. This type of locomotor response constitutes part of the body's adaptation strategy to prevent deviation from the habitual displacement of the COM [64]. Whether or not the runner is shod and the type of footwear where it is present (i.e., minimalist vs. conventional) are stimuli that can alter the Δy [64,172]. Kleg adjustments can compensate for the imposed perturbations caused by footwear condition and, thus, Δy, expressed by Kvert, remains unaffected [64,172]. This observation of increased Kleg and unaltered Kvert was notably unconfirmed by Divert and colleagues, who reported simultaneous increases in both Kvert and Kleg during the barefoot and minimalist running conditions [61,169]. The magnitude of the Kleg increase caused by barefoot running was proposed not to be adequate for maintaining unaltered Kvert. The Kvert increase arguably demonstrates the superiority of barefoot running in terms of energetic cost [169].

Ziliaskoudis and colleagues suggested that the COM might constitute greater vertical excursion when running barefoot when compared to shod running which could cause a total work production increase [25]. Supporting arguments reason that shoe sole geometry (thicker at the heel, thinner under the footballs) elevates the heel, compared to a more horizontal foot orientation relative to the ground at barefoot running [9]. Utz-Meagher and colleagues analysed precisely both barefoot and shod running [177]. The most commonly adopted pattern in barefoot running, FFS, also intensifies total work requirements by increasing joint excursions. The

amplified ankle joint plantarflexion before the instance of ground contact forces the footballs to touch down first [177]. A dorsiflexion movement then follows, enabling the heel to touch the ground before the reoccurrence of plantarflexion as the stance phase progresses [177]. However, shod running sees an absence of this initial lowering movement, due to the RFS pattern prevalence [177]. The escalated work requirements imposed by the different biomechanical characteristics between shod and barefoot running, although satisfied without metabolic penalty barefoot, do not occur during shod running [25]. The potential disadvantage when shod, as far as elastic energy storage and return is involved and compared with unshod, is not translated into an oxygen consumption (VO_2) increase due to lower total work demands [25]. Thus, although barefoot running is profitable from the perspective of net efficiency (that is, the total work production for a given amount of consumed oxygen), it does not appear to be advantageous in terms of running economy which is a distinct factor from net efficiency but still one with extensive influence on running performance [25].

Reviewing the above, it is also pertinent to refer to the recent development of light-weight materials. Wearing running shoes of such material, while limiting the disadvantage of added mass, also provides the energetically beneficial cushioning without contributing to total work production requirements [25]. It is plausible that technological evolutions in the fields of shoe design and materials' development, particularly in relation to the established relationship between running economy and stiffness, could compensate for the previously mentioned reduced ability of storage and return of elastic energy at shod running compared to barefoot running [25].

1.7.3. Surface –type and slope

Previous studies have shown Kleg modulation depends upon running service, with lower Kleg adopted on hard surfaces and higher Kleg on softer surfaces for the first step [64,178]. On lower-stiffness surfaces, runners decrease leg spring compression by increasing Kleg. This adjustment offsets the increased compression of the surface and keeps the path of a runner's COM the same regardless of its stiffness. As many biomechanical parameters are reliant upon the combined series stiffness of the runner and surface, modifying Kleg facilitates running in a similar manner on different surface stiffnesses [178]. Stride frequency, ground contact time, and peak GRF are all independent of this stiffness [178]. All of these observations are applicable to steady-state running on a continuous surface [64].

Little evidence is available on the interaction between slope gradients and stiffness. García-Pinillos and colleagues studied the impact of numerous factors on spatiotemporal parameters during running: slope gradient, athletic level, Kvert and Kleg [179]. Their findings on stiffness demonstrated that, relative to level running, Kvert increased with severe slopes (9-11%), whereas Kleg declined on moderate slopes (3-7%) regardless of the athletic level [179]. Alternate research concluded that Kvert was raised during uphill running whilst Kleg remained constant across the different gradients included in the study (-8 to 8%) [172]. The different methods in both studies may explain the slight differences between them. Whereas Lussiana required a pace of $10km \cdot h^{-1}$ over a range of slope gradients from -8 to 8%, García-Pinillos' study was executed at $12km \cdot h^{-1}$ using incremental slope gradients from 0-11%. Additionally, García-Pinillos and colleagues found notable correlation between Kleg and spatiotemporal parameters for level running, while Kvert was associated with spatiotemporal adaptations at more pronounced slope gradients (0-11%). The authors suggested that runners should engage greater force if they are to maintain velocity on steeper gradients resulting in increasing Kvert during running uphill [179].

1.7.4. Fatigue

The behaviour of the spring–mass model during a run to exhaustion is less apparent. In self-paced runs Kvert has been shown to decrease [67,180] with Kleg either decreasing (Hobara et al., 2010) or remaining unchanged [180]. The alternative approach has been to have participants run to exhaustion at a fixed exercise intensity. García-Pinillos and colleagues [181], found that in practised runners, Kleg decreased while Kvert remained consistent, which is supported by previous work [62,128]. However, Hunter and Smith found Kleg or Kvert did not vary in participants who lacked training [70]. Hayes and colleagues found that during a run to exhaustion both Kvert and Kleg decreased [126]. Though negligible difference in Kvert was observed, the change in Kleg was significant and of a moderate magnitude over the course of the run to exhaustion. They also found that the maintenance of Kleg held strong associations both with the distance time period covered by the run to exhaustion. What is more, participant ability to maintain Kleg was inversely proportionate to leg length change [126]. A non-significant decline of modest magnitude in Kvert reflected the findings of prior work on fixed velocity runs [62,128].

Regarding the relationship between velocity and stiffness, Enomoto and collegues suggested that stiffness adjustment to running speed is one of the key factors to keep pace in long-distance running [182]. These authors stated that to acquire running velocity effectively, a runner should run with suitable Kvert. Furthermore, they proposed that if a runner has high

Kvert even at low speed or keeps it despite the decrease in velocity, it might lead to fatigue and decrease in running velocity [182].

1.7.5. Velocity

Changes in lower-body stiffness over incremental-velocity protocol have already been reported [57,183]. The consequence of running at higher velocity are accepted to be increased step frequency, resulting in decreased CT, vertical displacement variation, as well as change in leg length [57]. Fluctuation in both Kvert and Kleg produced by increasing velocity correlates with spatiotemporal running gait characteristics [183]. Morin and colleagues determined that Kvert increases alongside increasing velocity while Kleg remains constant [57]. It is well-known that increasing running velocity produces an increase in step frequency, which results in decreased CT, Δy, as well as ΔL [129]. Fluctuation in both Kvert and Kleg produced by increasing velocity correlates with spatiotemporal running gait characteristics [183]. García-Pinillos and colleagues found that high-level runners displayed greater stride angle and FT at high velocity (18 $km \cdot h^{-1}$) and SL (at 14-16-18 $km \cdot h^{-1}$) although, their amateur counterparts exhibited higher SF at 11-16-18 $km \cdot h^{-1}$ [183]. Furthermore, amateur runners showed greater Kvert for all the velocities studied, whereas Kleg remained unchanged [183]. Although previous works found associations between running performance and lower-body stiffness [34,57,62,183-185], all the authors agreed on the lack of standardised methods to make comparisons related to spatiotemporal parameters and lower-body stiffness while running. It seems to be clear that Kvert increases as running velocity increases, while Kleg tends to remain unchanged.

1.7.6. Sex differences

Women demonstrate lower levels of active muscle stiffness than men at the same age, as shown by Wojtys [186]. Where controlled measurements of knee kinematics taken after mechanical perturbation during active flexion and extension exertions have recently been recorded and gender differences accounted for, women demonstrated less than 57% of the active muscle stiffness compared to males [187]. This may contribute to sex differences in musculoskeletal stability of the knee. It is unknown whether the lower stiffness measured in these controlled experiments translates to equivalent reductions in functional performance parameters such as Kleg during running and hopping. Active muscle stiffness contributes to Kleg and can be measured during functional tasks such as running and hopping have been reported [15,56,118]. The Kleg is attributed to the active muscle stiffness of the controlling joints [41] thereby affecting biomechanical stability. Granata and colleagues found that women exhibited lower Kleg than men in functional tasks while examining sex differences during two-legged

hopping [188]. Oscillation of the lighter female body mass necessitated the difference to facilitate equalling the hopping frequency of the heavier male subjects [188]. However, during preferred hopping conditions, no constraints were put upon the female participants to employ lower stiffness [188]. The women nevertheless consistently exhibited lower Kleg than the men, hopping at similar preferred frequencies and explanations for the mass-independent selection of this preference were proposed [188]. Similarly, Padua and colleagues exposed reduced Kvert in women, but the gender difference was eliminated once the body mass was normalised [83], which explains Granata and colleagues' suggestion in the aforementioned study; sex differences in Kvert during a functional hopping task can be plausibly explained by anthropometric differences [83,188]. Nevertheless, Padua and colleagues found different stiffness recruitment between men and women revealing that female quadriceps and soleus activity was significantly greater [83]. Whilst the recruitment strategy may, in principle, efficiently modulate Kvert, it also has the potential to compromise knee joint stability. Particularly for women, oestrogen, aside from its familiar role as a sex hormone, is also a crucial factor in the development, maturation, and aging of extragonadal tissues such as bone [189-191], muscle [192,193], and connective tissues [189,194]. There occurs natural variation in oestrogen secretion between young women, increasing 10- to 100-fold over the menstrual cycle [195]. As the concentration of oestrogen rises during the menstrual cycle, so knee laxity rises, hence joint laxity has been found to be cyclical in nature [196-198]. A change in knee laxity from 13.35 ± 2.53 mm during the follicular phase to 14.43 ± 2.60 mm during ovulation resulted from a 17% reduction in knee stiffness during the ovulatory phase [199]. Since the properties of ligaments and tendons vary across the menstrual cycle, it should be considered while testing stiffness in women.

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2. HIPOTHESIS

Regarding the information shown in the previous section, it has been hypothesised that the OptoGait photoelectric cell system constitutes a reliable tool in treadmill running to gauge spatiotemporal parameters and lower-limb stiffness. Additionally, it is hypothesised that SSC and lower-limb stiffness react differently within the same runners while jumping and running due to the specificity principle behind the tasks. Ultimately, it has also been hypothesised that footwear condition, FSP, and SF influence on lower-limb stiffness while running making Kvert and Kleg fluctuate according to the manipulation of such factors.

3. AIMS

3.1. General aim

The present PhD Thesis aims to determine the effect of various influential factors on lower-limb stiffness in long-distance running in healthy adults.

3.2. Specific aims

Specifically, this PhD Thesis aims to

- identify influencing factors on lower-limb stiffness while running from both injury prevention and performance perspectives;
- determine the test-retest reliability of the OptoGait™ photoelectric cell system for spatiotemporal parameters and lower-limb stiffness analysis at a comfortable velocity;
- determine the relationship between reactivity and lower-limb stiffness in amateur endurance runners during jumping and running as well as sex differences;
- analyse the influence of footwear condition, foot-strike pattern, and step frequency on spatiotemporal parameters and both vertical and leg stiffness while running.

4. MATERIAL AND METHODS

Throughout this section, materials and methods used to pursue this PhD Thesis are described. First, variables and research tools shared amongst all the studies are explained whereas, Table 1 summarises all the material used and methods implemented over the development of each particular study of the present PhD Thesis.

The spatiotemporal parameters of CT, FT, SL, %CT, and %FT were measured using OptoGait™ photoelectric cell system (Microgate, Bolzano, Italy), which has been previously validated for the assessment of racewalking spatiotemporal parameters [1]. The system calibration was done by the manufacturer recommendations and consisted of two-transmitting receiving bars placed parallel to one another and, for all the studies implemented in the pursue of this PhD Thesis, set on the treadmill surface (HP cosmos Pulsar 4P; HP cosmos Sports & Medical, Gmbh, 138 Nußdorf, Germany) (Figure 1). The OptoGait system was linked via a USB cable to a laptop and the manufacturer software used (Version 1.12.1.0, Microgate, Bolzano, Italy). Filter parameters GAitR-In and GAitR-Out were set at 0_0 [2]. Data was collected at a sampling of 1,000 Hz, encrypted, and stored securely. Following Brown recommendations [3], limb dominance was not considered. The spatiotemporal parameters analysed in the present PhD Thesis has already been described accurately by previous research [4]:

- CT (s): time spent in contact with the ground on each step by one foot from initial contact to the very early stage of toes lifting off the ground.
- FT (s): time spent from the early stage of toes lifting off to the very early stage of initial contact in the next footfall.
- SF (spm): amount of ground contact events in a minute.
- Percentage of CT (%CT) and FT (%FT) over the step cycle.

Kvert and Kleg, previously described in this PhD Thesis, have been determined by Morin's sine-wave method [5]. As already mentioned, in order to estimate Kvert and Kleg values, Morin's approach requires us to gather a small quantity of information (CT, FT, L, v, and m). Indeed, there is a small difference (0.67-6.93%) between stiffness when calculated using platform and the sine-wave methods [5]. Also, it has been reported that the sine-wave approach is suitable for Kvert and Kleg analysis for intra and inter-day designs (ICCs = 0.86-0.99) [6].

Table 1. Methods used in the current PhD Thesis

Study	Study design	Participants	Protocol	Outcome measures
1. Test–retest reliability of the OptoGait system for the analysis of spatiotemporal running gait parameters and lower body stiffness in healthy adults	Unilateral cross-over	n=31 (18 males, 13 females) age: 34.42 ± 9.26 years; height: 171.54 ± 9.15 cm; body mass: 66.63 ±11.3 kg	Two-session protocol 3-min treadmill run at 12 km·h^{-1}, 1-week washout period, 3-min treadmill run at 12 km·h^{-1}	CT, FT, SL, SF, Kvert, Kleg
2. Is there a relationship between reactivity and stiffness in amateur endurance runners? A comparative analysis between sexes	Unilateral cross-over	n=19 (14 males, 5 females) age: 27.9±6.4 years; height: 172.7±7.4 cm; body mass: 66.2±10.5 kg; 10-km time < 50 min	6 DJ (3x20cm height: 3x30cm height), 3-min treadmill run at 12 km·h^{-1} shod and barefoot	Kvert, Kleg, RSI, foot arch stiffness
3. How do footwear, foot-strike pattern and step frequency influence on spatiotemporal parameters and lower-body stiffness in endurance running?	Unilateral cross-over	n=31 (18 males and 13 females) age: 34.42 91 ± 9.26 years; height: 171.54 ± 9.15 cm; body mass: 66.63 ± 11.3 kg; 10-Km time: 48.46 ± 3.85 min)	Two-session protocol 3-min shod treadmill run at 12 km·h^{-1}, SF alteration (150-160-170-180-190), 1-week washout period, 3-min barefoot treadmill run at 12 km·h^{-1}, SF alteration	%CT, %FT, Kvert, Kleg

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5. RESULTS

This section is presented in the final version in which each study has been submitted to the different scientific journals. Tables and figures are provided within each study as well as a detailed description of the research outcomes in the corresponding results section. Additionally, the main findings of these studies are summarised at the end of this section (Table 2).

Study 1. Test-retest reliability of the OptoGait system for the analysis of spatiotemporal running gait parameters and lower-body stiffness in healthy adults

Jaén-Carrillo, D., García-Pinillos, F., Cartón-Llorente, A., Almenar-Arasanz, A. J., Bustillo-Pelayo, J. A., & Roche-Seruendo, L. E. (2020). Test–retest reliability of the OptoGait system for the analysis of spatiotemporal running gait parameters and lower body stiffness in healthy adults. *Proceedings of the Institution of Mechanical Engineers, Part P: Journal of Sports Engineering and Technology,* 1754337119898353. (Appendix 1)

Test-retest reliability of the OptoGait system for the analysis of spatiotemporal running gait parameters and lower-body stiffness in healthy adults

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ABSTRACT

Despite the widespread use of the OptoGait photoelectric cell system for the analysis of running spatiotemporal parameters, its reliability has not been proved. Consequently, this study intends to determine the test-retest reliability of the system when applied to treadmill running spatiotemporal parameters and lower-body stiffness at a constant velocity. Amateur endurance runners (n=31; age: 34.42 ± 9.26 years; height: 171.54 ± 9.15 cm; body mass: 66.63 ± 11.3 kg) voluntarily consented to participate in this study. Data for each participant was recorded twice per session across two testing sessions. The intra-session and inter-session reliability of spatiotemporal parameters and lower-body stiffness were determined through test-retest analysis. Although mean comparisons revealed significant differences between measurements in spatiotemporal running gait characteristics and lower-body stiffness for intra-session ($p < 0.05$ in all parameters), the effect size was always small (< 0.4). Moreover, the relationship between measurements was very large for spatiotemporal parameters and lower-body stiffness ($r > 0.7$). The ICCs revealed an almost perfect correlation between measurements (ICCs > 0.81), except Kleg with substantial reliability (ICC = 0.788). The inter-session reliability revealed some significant differences in CT ($p= 0.009$) and Kleg ($p=0.013$), although Cohen´s d indicated small ES (< 0.31). The relationship between sessions was very large for spatiotemporal parameters and lower-body stiffness ($r > 0.8$), and the ICCs revealed an almost perfect inter-session association (ICCs > 0.881). The results found here show that the OptoGait system can be used confidently for running spatiotemporal parameters analysis and lower-body stiffness at a constant velocity for healthy adults.

Keywords: Reliability, testing, OptoGait, running, stiffness

INTRODUCTION

Running is an enduringly popular pursuit. Benefits include improved cardiovascular function and mental health, stress relief, and enjoyment (1-4). When animals run, they bounce along the ground. Such movement is facilitated by a system of musculoskeletal springs, comprised of muscles, tendons and ligaments which store elastic energy when stretched and release it when recoiled (5-7). During running, this complex musculoskeletal system behaves much like a single linear spring (the 'leg spring') (8). In fact, a simple spring-mass model consisting of a single linear leg spring and a mass equivalent to that of the animal has been shown to describe and predict the mechanics of running remarkably well (9-14). As vertical stiffness (Kvert) and leg stiffness (Kleg) influence the regulation of both spatiotemporal and kinematic variables, they are usually used while identifying these characteristics in individual runners. The Kvert (kN/m) is the ratio of maximal force to the vertical displacement of the centre of mass as its lowest point is reached (i.e., the middle of the stance phase). Similarly, Kleg (kN/m) is defined as the ratio of the maximal force in the spring to the maximum leg compression at the middle of the stance phase (8, 15).

While some previous studies described the influence of contact time (CT) and Kleg on both performance and running economy (16, 17), others have not demonstrated this influence (18, 19). Limitations of the methods in use for running biomechanics analysis might be the main reason for this difference. The drawbacks to commercially available tools for such analysis include limited accessibility, high cost, sensory fragility and operating complexity, and they are mainly employed in research rather than clinical settings. It has been shown that high-speed video analysis is a reliable and valid method to measure running kinematics (20), as well as 3-D motion capture system – considered as a 'gold standard'. However, running kinematics analysis using the systems mentioned above requires, among others, highly-trained users for proper data collection, as well as data analysis. Floor-level, high-density photoelectric cells (OptoGait, Microgate, Bolzano, Italy), which are portable and allow quantification of spatiotemporal gait parameters on most flat surfaces, are used for clinical purposes (21).

Although previous research concerning the OptoGait™ system has considered its reliability in assessing spatiotemporal walking and racewalking gait variables (21-23), measuring spatiotemporal gait characteristics during running by implementing an incremental speed protocol (24, 25) and calculating both Kvert and Kleg while running on a treadmill with different slope gradients (26), the system reliability for the analysis of running gait

spatiotemporal parameters, as well as lower-body stiffness, is still unknown. Therefore, the aim of this study is to analyse the test-retest reliability of spatiotemporal gait characteristics and lower-body stiffness while running on a treadmill at a constant velocity by comparing data intra-session and inter-sessions.

METHODS

An observational study, aligned with STROBE guidelines (27), was conducted for accuracy diagnosis of running gait. The running spatiotemporal parameters of CT, flight time (FT), step length (SL), and step frequency (SF) were analysed, as well as both Kvert and Kleg. This study was approved by the ethics committee at the University San Jorge (Zaragoza, Spain) (Appendix 1).

Participants

A sample of 31 healthy subjects, 18 men and 13 women, (age: 34.42 ± 9.26 years; height: 171.54 ± 9.15 cm; body mass: 66.63 ± 11.3 kg), who were accustomed to running on a treadmill and able to run 10 km in 50 – 60 minutes, voluntarily participated in this study. Informed consent (Appendix 2), which complied with the standards of the Declaration of Helsinki of the World Medical Association, was obtained from all participants prior to the study. Subjects who reported musculoskeletal injuries sustained within the previous six months or suffered from any other impairment that might affect their running gait were excluded from the study. Consequently, participants were free from cardiovascular, neurologic, or musculoskeletal conditions and familiar with running on a treadmill. The recruitment was done amongst sport sciences students.

Procedures

This study was executed in the biomechanics laboratory at the University San Jorge across two different sessions. Participants performed the same protocol under the same conditions. They were instructed by a researcher and completed the entire protocol running on a treadmill with an established inclination of 0%. Subjects started warming up at a speed of 8 km/h, increasing it freely over the course of eight minutes ultimately reaching 12 km/h because previous studies (28, 29) have shown that accommodation to treadmill running on human locomotion takes approximately 6-8 minutes. After the warm-up, participants ran at a speed of 12 km/h for three minutes during which time data were recorded for analysis. Subsequently, subjects ran for five minutes at a self-selected speed. Then they ran again for three more minutes at 12 km/h with

data recorded for analysis. Subjects left the biomechanics laboratory after completing the running. One week later, subjects returned and repeated the same procedure under the same conditions. Subjects were instructed to continue their regular training but were asked to avoid competitions and high-intensity activities 24 hours for the study. All the steps occurred in the sensor area during analysis. Besser et al. showed that recording 6–8 strides was adequate to acquire representative data for healthy adults (defined as 95% confidence intervals within 5% of error) (30).

Both body mass (kg) and height (cm) for each participant were found using a weighting scale (Tanita BC-601; TANITA Corporation, Maeno-Cho, Itabashi-ku, Tokyo, Japan) and a precision stadiometer (SECA 222; SECA Corp., Hamburg, Germany), respectively. Participants wore only underwear during these measures. The leg length (L) of each participant was found in accordance with Winter's (31) anthropometric equations as shown in Eq. (12):

$$L = 0.53h \tag{12}$$

where h stands for the participant's height (m).

The running spatiotemporal parameters of CT (s), FT (s), SL (cm) and SF (spm) were measured using the OptoGait Photoelectric Cell system (OptoGait, Microgate, Bolzano, Italy) – previously validated for the evaluation of spatiotemporal features of the gait in young adults (23). The OptoGait system calibration was done by the manufacturer and consisted of two transmitting-receiving bars placed parallel to one another, set on the treadmill surface for this study (HP cosmos Pulsar 4P; HP cosmos Sports & Medical, Gmbh, Nußdorf, Germany) (Fig.

Figure 5. Location of the OptoGait system on a treadmill

4). The OptoGait system was linked via a USB cable to a personal computer and the manufacturer's software was used (Version 1.12.1.0, Microgate, Bolzano, Italy). In order to minimise the systematic bias, the filter parameters GAitR-In and GAitR-Out were both set at 0_0 (23, 32). The data was extracted at a sampling frequency of 1,000 Hz, encrypted and stored on a computer. According to Brown et al. (33), limb dominance was not considered.

This study employs a procedure developed by Morin et al. to determine lower-body stiffness (15). Kvert (kN/m) determines the general level of stiffness in the body by finding the ground reaction force and vertical displacement of the centre of mass relationship, whereas Kleg (kN/m) shows the stiffness in just the lower part of the body (feet, ankles and hip joints) and gives the ratio between the ground reaction force and the deformation in leg length (15). Morin's approach is useful because it only requires gathering a relatively small amount of information (CT, FT, leg length, speed, and body mass) to calculate the runner's approximate Kvert and Kleg. Indeed, these authors have demonstrated that there is a small difference (0.67-6.93%) between stiffness when calculated using the sine-wave and platform methods (15). For their part, Pappas et al. confirmed that the sine-wave method could be used to accurately measure Kvert and Kleg for intra and inter-day designs with ICCs between 0.86-0.99 (34).

Data analysis

Descriptive statistics are represented as mean (±SD). Tests of normal distribution and homogeneity by the Kolmogorov-Smirnov and Levene's test, respectively, were conducted on all data before analysis. A mean comparison analysis (T-test) was conducted between variables from both measurements (i.e., intra-session) and from both days (i.e. inter-session). The magnitude of the differences was interpreted using Cohen's d effect size (ES) (10). Effect sizes are reported as: trivial (<0.2), small (0.2-0.49), medium (0.5-0.79), and large (≥0.8) (10). The relationship and association of variables from different measurements (i.e. intra-session) and from different testing days (i.e. inter-session) were quantified through the Pearson correlation coefficient (r) and the intraclass correlation coefficient (ICC). The following criteria were adopted to interpret the magnitude of correlations between measurement variables: <0.1 (trivial), 0.1–0.29 (small), 0.3–0.49 (moderate), 0.5–0.69 (large), 0.7–0.89 (very large) and 0.9–1.0 (almost perfect) (11). Based on the characteristics of this experimental design and following the guidelines reported by Koo and Li (35) (12), the authors decided to conduct a "two-way random-effects" model (ICC [2,k]), "mean of measurements" type, and "absolute" definition for the ICC measurement. The interpretation of the ICC was based on the

benchmarks reported by a previous study (13): ICC < 0 reflects 'poor', 0-0.20 'slight', 0.21-0.40 'fair', 0.41-0.60 'moderate', 0.61-0.80 'substantial', and > 0.81 'almost perfect' reliability. The Bland-Altman (14) limits of agreement method (mean difference ± 1.96 SD) was used to analyse differences in spatiotemporal features and lower-body stiffness between measurements (i.e., intra-session) and between testing sessions (i.e., inter-session). Heteroscedasticity of error was defined as an r2>0.1. All the statistical analyses have been executed following the suggestions done by Atkinson and Nevill for assessing reliability (36). The level of significance used was p<0.05. Data analysis was performed using the SPSS (version 21, SPSS Inc., Chicago, Il.).

RESULTS

The intra-session reliability of spatiotemporal parameters and lower-body stiffness were determined through test-retest analysis (Table 3). Despite mean comparisons (i.e., measurement 1 vs. measurement 2) which revealed significant differences between measurements in spatiotemporal gait characteristics and lower-body stiffness (p < 0.05 in all parameters), the effect size was always small (< 0.4). Additionally, the relationship between measurements was very large for spatiotemporal parameters and lower-body stiffness (r > 0.7). The ICCs also revealed an almost perfect association between measurements (ICCs > 0.81), except from Kleg with substantial reliability (ICC = 0.788).

Table 2. Intra-session reliability of spatiotemporal parameters and lower-limb stiffness

Variable	Measurement 1 (±SD)	Measurement 2 (±SD)	P-value (Cohen´s d)	Pearson coefficient (r)	ICC (95% CI)
CT (s)	0.274 (0.020)	0.280 (0.017)	0.006 (0.323)	0.811***	0.865 (0.670-0.940)
FT (s)	0.085 (0.021)	0.081 (0.019)	0.055 (0.199)	0.841***	0.904 (0.795-0.954)
SL (cm)	120.40 (5.48)	121.45 (5.08)	0.012 (0.209)	0.919***	0.948 (0.874-0.977)
SF (spm)	166.55 (7.60)	164.91 (7.30)	0.007 (0.214)	0.911***	0.943 (0.854-0.975)
Kvert (kN/m)	22.19 (3.41)	21.71 (3.26)	0.049 (0.148)	0.924***	0.956 (0.905-0.979)
Kleg (kN/m)	7.33 (1.01)	6.96 (0.89)	0.009 (0.389)	0.700***	0.788 (0.521-0.902)

CT: contact time; FT: flight time; SL: step length; SF: step frequency; Kvert: vertical stiffness; Kleg: leg stiffness; ICC: intraclass correlation coefficient; CI: confidence interval; SD: Standard deviation
*** denotes p < 0.05

The pairwise comparisons between testing days (i.e., inter-session reliability) revealed some significant differences in CT (p= 0.009) and Kleg (p=0.013), although Cohen´s d indicated small ES (< 0.31) (Table 4). The relationship between sessions was very large for spatiotemporal parameters and lower-body stiffness (r > 0.8), and the ICCs revealed an almost perfect inter-session association (ICCs > 0.881).

Table 3. Inter-session reliability of spatiotemporal parameters and lower-limb stiffness

Variable	Day 1 (±SD)	Day 2 (±SD)	P-value (Cohen´s d)	Pearson coefficient (r)	ICC (95% CI)
CT (s)	0.274 (0.020)	0.268 (0.018)	0.009 (0.315)	0.850***	0.900 (0.743-0.957)
FT (s)	0.085 (0.021)	0.089 (0.022)	0.446 (0.186)	0.814***	0.894 (0.771-0.951)
SL (cm)	120.40 (5.48)	120.59 (5.57)	0.649 (0.07)	0.843***	0.916 (0.819-0.961)
SF (spm)	166.55 (7.60)	166.12 (7.85)	0.799 (0.052)	0.852***	0.921 (0.828-0.963)
Kvert (kN/m)	22.19 (3.41)	22.40 (4.23)	0.064 (0.05)	0.896***	0.896 (0.770-0.952)
Kleg (kN/m)	7.33 (1.01)	7.60 (1.25)	0.013 (0.238)	0.833***	0.881 (0.709-0.948)

CT: contact time; FT: flight time; SL: step length; SF: step frequency; Kvert: vertical stiffness; Kleg: leg stiffness; ICC: intraclass correlation coefficient; CI: confidence interval; SD: Standard deviation
*** denotes $p < 0.05$

Through Bland-Altman plots, Figs. 5 and 6 show the intra-session differences between the measurements (systematic bias and random error) and the degree of agreement (95% limits of agreement). Small biases and errors were observed in spatiotemporal parameters (CT: -0.01 ± 0.01 s; FT: 0.00 ± 0.01 s; SL: -1.04 ± 2.16 cm; SF: 1.63 ± 3.15 spm) (Fig. 5) and vertical and leg stiffness (Kvert: 0.48 ± 1.31 kN/m; Kleg: 0.37 ± 0.74 kN/m) (Fig. 6). No heteroscedasticity of error was found in any variable (r2 < 0.1).

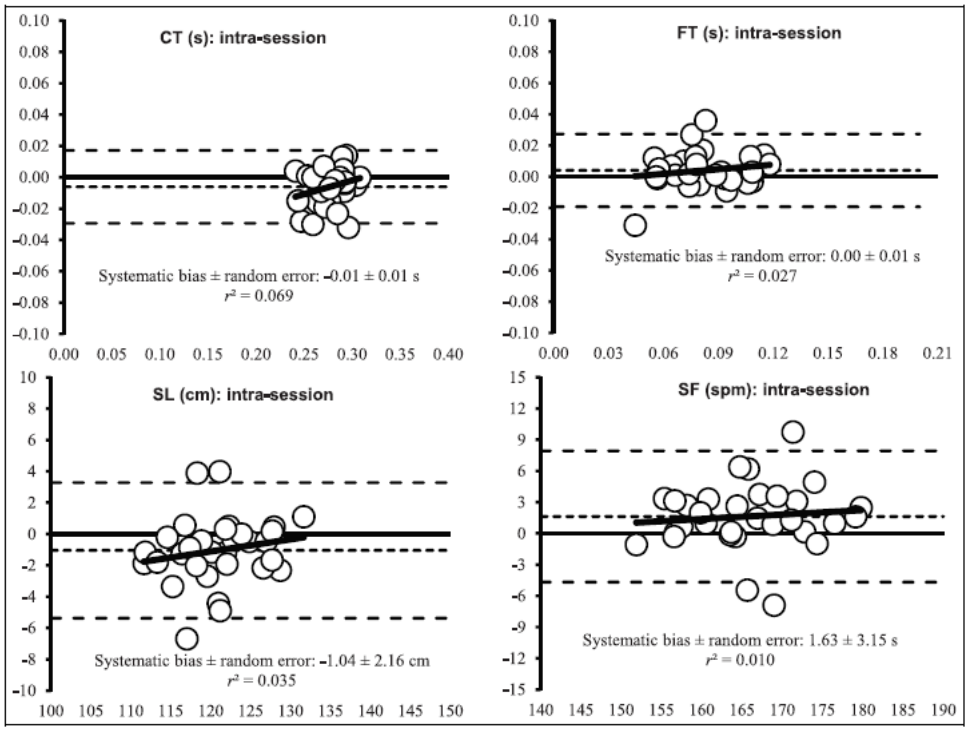

Figure 6.Intra-session differences between the measurements (systematic bias and random error) and the degree of agreement (95% limits of agreement) for CT, FT, SL, and SF.

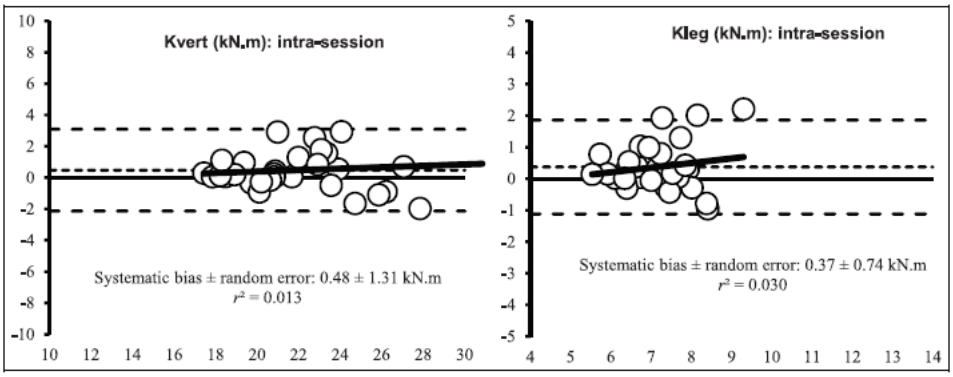

Figure 7. Intra-session differences between the measurements (systematic bias and random error) and the degree of agreement (95% limits of agreement) for Kvert and Kleg.

Bland-Altman plots also show the inter-sessions differences in the measured variables and the degree of agreement between the two measurements. Small systematic biases and random errors were reported for spatiotemporal parameters (CT: 0.01 ± 0.01 s; FT: 0.00 ± 0.01 s; SL: 0.27 ± 3.05 cm; SF: -0.20 ± 4.15 spm) with no heteroscedasticity of error (r2 < 0.1) (Fig. 7). As for vertical and leg stiffness, despite biases and errors, they were small (Kvert: -0.78 ± 2.12 kN/m; Kleg: -0.35 ± 0.69 kN/m), while heteroscedasticity of error was found in both variables (Kvert: r2 = 0.472; Kleg: r2 = 0.107) (Fig. 8).

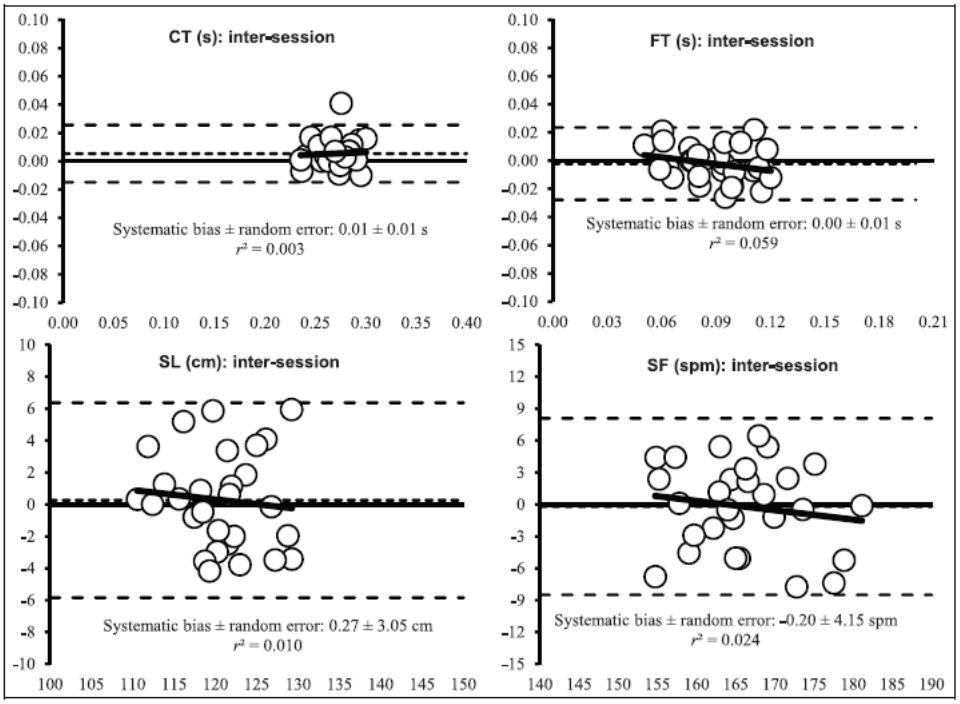

Figure 8. Inter-session differences between measurements and degree of agreement for CT, FT, SL, and SF.

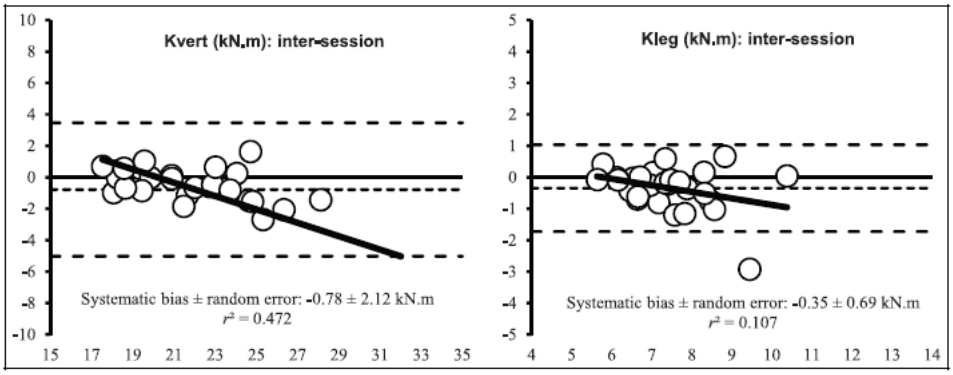

Figure 9. Inter-session differences between measurements and degree of agreement for Kvert and Kleg.

DISCUSSION

This study aimed to analyse the test-retest reliability (i.e., intra- and inter-session) of the OptoGait system for the acquisition of spatiotemporal gait characteristics and lower-body stiffness while running on a treadmill. The study tested 31 healthy adults to demonstrate the reliability of the OptoGait system, while acquiring the running spatiotemporal parameters of FT, CT, SL, and SF, as well as both Kvert and Kleg. The results indicate that spatiotemporal parameters and lower-body stiffness during running were reliable in both intra- and inter-session contexts. Nevertheless, the Bland-Altman analysis provides insights into the systematic differences between the measurements. None of the measured variables reported heteroscedasticity of error, except vertical and leg stiffness in the session 1 vs. session 2 comparison (i.e., inter-session reliability). The results reinforce the intra- and inter-session reliability data for spatiotemporal parameters and intra-session reliability for lower-body stiffness, but it also warns about the lack of stability of the Kvert and Kleg variance.

Reliability is essential for a running gait analysis system to guarantee that differences in running gait performance are related to gait changes as opposed to errors in data collection. The current findings are similar to previously reported results regarding the spatiotemporal parameters for healthy adults (21, 22). Whilst Gomez Bernal et al. tested the reliability of the OptoGait system for spatiotemporal parameters analysis while walking on a treadmill and Lee et al. asked their participants to walk three times on a walkway at a comfortable velocity, the current study shows the test-retest reliability of the OptoGait system for treadmill running spatiotemporal parameters analysis. Compared to previous studies where running spatiotemporal parameters were measured using the OptoGait system (25), an incremental velocity protocol (10 to 20 km/h) was implemented in various studies to measure running

spatiotemporal parameters in contrast to the current study, where a constant velocity (12 km/h) was established during data collection to examine the test-retest reliability of the OptoGait system for treadmill running spatiotemporal parameters. Due to the lack of available information regarding the use of the OptoGait system for spatiotemporal parameters while running at a constant velocity, it makes comparison to other studies difficult, thus underscoring the importance of this study.

It has been demonstrated that the value of Kvert is always higher than Kleg in locomotion since variations in leg length surpass those of the centre of mass (15, 34). Despite Kvert and Kleg being derived from analogous mechanical concepts, they are not equivalent and adapt differently to fluctuations in running conditions (8, 15). Hence, the evaluation of both Kvert and Kleg is justified. The findings reported here of the intra-session trials correlate perfectly with those found by Pappas et al. (34) as shown respectively in the following parentheses regarding ICCs for FT (0.904 and 0.970), SL (0.948 and 0.925), SF (0.943 and 0.932), and Kvert (0.956 and 0.972) and differ slightly for CT (0.865 and 0.986) and Kleg (0.788 and 0.982). In regard to the findings of the inter-session trials, the results found in the current study are very similar to Pappas et al.'s results as shown respectively in the following parentheses regarding ICCs for CT (0.900 and 0.925), FT (0.894 and 0.902), SL (0.916 and 0.860), SF (0.921 and 0.863), Kvert (0.896 and 0.922), and Kleg (0.881 and 0.873). The slight differences between both studies might be related to differences in methods. While Pappas et al. only included male participants, the participants for the current study included both male and female runners. Moreover, Pappas and colleagues recorded three rounds of 30 seconds at 16 km/h for each participant compared to the current study where data for each participant was recorded once over three minutes at a constant velocity of 12 km/h. It has been demonstrated that longer recording periods return smaller systematic bias and random errors, as well as narrower limits of agreement regarding step variability (37).

Although the current study sheds some light on the use of the OptoGait system as a reliable tool for the analysis of running spatiotemporal parameters, some limitations must be considered. On the one hand, the laboratory scene should be considered while interpreting these findings; nevertheless, participants were accustomed to running on a treadmill. On the other hand, although Morin's approach (15) shows good efficacy and accuracy for the analysis of lower-body stiffness, it is not a direct method. The strong reliability of the OptoGait system demonstrated by the current results will provide future researchers enough evidence to use this photoelectric system for the accuracy analysis of running spatiotemporal parameters and lower-

body stiffness. Since healthy adults have been evaluated in this study, future research work should consider the assessment of the system for different ages and population suffering from musculoskeletal pathologies.

CONCLUSION

The current study shows that the OptoGait system performs reliable evaluation for running spatiotemporal parameters analysis and lower-body stiffness at a constant velocity for healthy adults. The findings reported here might have a high importance for sport scientists and clinicians working on both running gait retraining and improvement. The user-friendliness of the OptoGait system and its proved reliability for running spatiotemporal parameters analysis provide coaches and clinicians a trustworthy instrument to make judgements regarding the degree of change related to the normal variability of measuring between trials or sessions, especially for early identification of running pathologies.

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Study 2. Is there a relationship between reactivity and stiffness in amateur endurance runners? A comparative analysis between sexes

Under review in Sports Biomechanics

ABSTRACT

This study examined the relationship between reactive strength index and lower-limb stiffness in amateur endurance runners during jumping and running. Nineteen runners (14 males, 5 females; age: 27.9 ± 6.4 years; height: 172.7 ± 7.4 cm; body mass: 66.2 ± 10.5 kg; 10-km time < 50 min) volunteered to participate. Kinematic and kinetic data were recorded over one session. First, participants executed 6 drop jumps (3 x 20cm; 3 x 30cm height) and the highest jump for each height was analysed. Then, runners performed two 3-min treadmill running trials at 12 $km \cdot h^{-1}$, shod and barefoot conditions. Shapiro-Wilk's test confirmed normal distribution data. Repeated measures ANOVAs resulted in significant sex differences for arch and vertical stiffness at shod running, exhibiting men greater values ($p < 0.05$). ANCOVA resulted in significant sex differences for reactive strength index, showing men greater values, and 30-cm drop jump performance ($p < 0.05$). Cohen's d was used to interpret effect size. The results indicate that the spring-mass model reacts differently to tasks based on their specificity principle. Additionally, sex-related differences must be considered when assessing the stretch-shortening cycle.

Key Words: jumping, long-distance running, sex differences, spring-mass model, strength

INTRODUCTION

During running, the leg function resembles the behaviour of a spring which compresses and decompresses continually (1), being the body mass responsible for such leg-spring compression (2). Mechanical energy is stored over the leg-spring compression, represented by the eccentric phase of stance, whereas the concentric phase of stance releases the stored energy as elastic energy (2). Lower-limb stiffness (3) and the stretch-shortening cycle (SSC) (4) are the two most important neuromuscular elements linked to elastic energy use.

Testing stiffness in running gait includes the highest specificity for runners and may be calculated at the most suitable speed for the athlete (i.e., race pace) (5). Vertical stiffness (Kvert) and leg stiffness (Kleg) influence the regulation of both running spatiotemporal and kinematic variables. While Kvert describes the ratio of maximal force to the vertical displacement of the centre of mass (COM) when its lowest point is reached, Kleg describes the mechanical behaviour of the leg's structural components (i.e., muscles, tendons, and ligaments) (6-8). Increases in Kvert and Kleg are connected to both increased intensity and improved performance of a particular task (5).

Sex differences in lower-limb stiffness have already been reported during jump landings (9) and hopping tasks (10). Considering the spring-mass model, lower-limb stiffness increases with body mass (11), therefore, the lighter body mass of women may account for the sex differences found. A recent study compared lower-limb stiffness between ballet dancers and team sport athletes using single-legged DJs from a 30-cm box (12). The authors found that men showed greater lower-limb stiffness than their female counterparts, displaying lower COM vertical displacement per height and greater ground reaction force per body mass (12), being consistent with previous studies (9). Also, it has been suggested that stiffness should be evaluated under fast SSC conditions where an initial impact phase is involved (i.e., drop jump [DJ]) (5).

The SSC, distinguished as slow or fast (contact time [CT] ≤ 250ms) (13), is a natural type of muscle function in which muscle is stretched immediately prior to shortening and it is observed in various everyday activities such as running and jumping (14). This eccentric/concentric coupling of muscular contraction produces a more powerful contraction than that which would result from a purely concentric action alone (14). The mechanisms underpinning any SSC activity can be determined by the demands of the SSC criterion task (15).

Reactive strength index (RSI), defined as the ability to switch rapidly from an eccentric to a concentric contraction (16), can be measured from the ratio of flight time (FT) and CT by DJ (17). The demands of a task, dictated by the principle of specificity, will determine the type of SSC used (15), and therefore, the RSI values. It has been previously shown that RSI vary between men and women, showing men greater values than women over different sports (18, 19). Suchomel and colleagues found differences in RSI values between male and female soccer players at the same competition level, but no differences were reported between tennis players under the same conditions (18). The correlations between the RSI and stiffness using DJs at different heights have been investigated (20). A positive correlation between RSI and Kvert was found, meaning that greater stiffness values were associated with greater RSI (20). As reported in the García-Pinillos and colleagues' study, Kvert values over the DJs remained constant across drop heights (20). This finding might be expected as both COM displacement and peak ground reaction force are used for Kvert calculation and both increased during DJs simultaneously with drop height. Similarly, Ferris and colleagues suggested that people tend to maintain Kvert unchanged to keep movement mechanics stable (21, 22).

To the authors' knowledge, the connections between RSI and lower-limb stiffness across different specific tasks such as running and jumping within the same group remain unknown. Hence, this study is aimed at examining the relationship between RSI and lower-limb stiffness during jumping and running across the same participants, as well as identifying possible sex differences. We hypothesised that RSI and lower-limb stiffness react differently within the same runners while jumping and running due to the different tasks' specificity. Additionally, we hypothesised that females display lower values for both RSI and lower-limb stiffness than males.

METHODS

This study aimed at analysing the relationship between lower-limb stiffness and reactivity in amateur endurance runners as well as to highlight the influence of sex on that relationship. Both Kvert and Kleg while shod and barefoot running at a constant speed were found. Additionally, RSI was calculated for every participant, as well as arch stiffness. In order to shed some light on the relationship between these variables, and taking sex differences into consideration, a unilateral crossover design was executed. The study was approved by the Institutional Review Board (University San Jorge, Zaragoza, Spain) (Appendix 3).

Participants

A group of nineteen amateur endurance runners voluntarily participated in this study (14 men and 5 women; age: 27.9±6.4 years; height: 172.7±7.4 cm; body mass: 66.2±10.5 kg). All participants were habitually shod runners and met the inclusion criteria: (1) older than 18 years old, (2) able to run 10 km in less than 50 minutes, (3) at least two running sessions weekly, and (4) not suffering from any injury in the last 6 months before the data collection. After receiving detailed information on the objectives and procedures of the study, each participant signed an informed consent form in order to participate (Appendix 4), which complied with the ethical standards of the World Medical Association's Declaration of Helsinki (2013). It was made clear that the participants were free to withdraw from the study at any time. Sport sciences students were recruited for the study.

Procedures

Data was collected over only one session in the biomechanics laboratory of the University San Jorge during March and April 2019. Participants ran shod over 3 minutes for the first running trial and barefoot during 3 minutes for the second running trial. Both runs were completed on a motorised treadmill with a slope of 0% (HP cosmos Pulsar 4P; HP cosmos Sports & Medical, Gmbh, Nußdorf, Germany). The following procedure under the same conditions, and instructed by a researcher, was performed by every participant.

Before the start of the testing session, the participants completed a dynamic warm-up protocol that consisted of movement preparation (squatting, lunging, and hinging), 5-min stationary cycling, dynamic stretching, running drills consisting of skipping, counter-movement jump (CMJ), CMJ with bounce, and ankle jumps. It is suggested that this type of warm-up stimulates greater jumping performance (23). Each participant performed 3 maximal jumps at each of the 2 drop-jump (DJ) heights (20 and 30 cm box; 6 jumps in total) and the best performance was considered for analysis. The landing zone was established between two transmitting-receiving bars belonging to photoelectric cell system (OptoGait, Microgate, Bolzano, Italy) –previously validated to measure vertical jump height (24). Measurements of FT (ms) and CT (ms) were recorded, and their ratio (RSI) was found. RSI was found to be a reliable and valid indicator of explosive performance (18, 25). Participants had 1-min rest between jumps and a 3-min recovery between DJ heights (26). To start, participants were asked to 'step out' from the box, keeping their hands on their hips to minimise arm movement, and 'to jump as high and as fast as possible' on landing (27). Each jump was analysed carefully and considered unacceptable

in case that either the participants' legs were not fully extended during the flight or they jumped forward off the landing zone.

Immediately after, an 8-min accommodation program (28) was implemented by increasing speed by 1 $km \cdot h^{-1}$ every minute from 8 to 12 $km \cdot h^{-1}$. Then, participants ran shod at a speed of 12 $km \cdot h^{-1}$ for 3 minutes, and 6-8 strides were analysed since it has been suggested as adequate to acquire representative data in healthy adults (95% confidence intervals within 5% of error) (29). Thereafter, participants ran barefoot at 12 $km \cdot h^{-1}$ for another 3 minutes. Data were recorded during both running trials for subsequent data analysis.

Materials and testing

As participants entered the laboratory weight (kg) and height (cm) were determined using a weighing scale (Tanita BC-601; TANITA Corp., Maeno-Cho, Itabashi-ku, Tokyo, Japan) and a stadiometer (SECA 222; SECA Corp., Hamburg, Germany).

Total foot length (FL), truncated foot length (TFL), and arch height were defined following Butler et al., who reported high intra and interrater reliability (30). First, linear dimensions of the unloaded right foot placed on an osteometric board were measured using sliding digital callipers with participants seated in a height-adjustable chair keeping their knees and hips under an alignment of 90° (31). Feet, positioned 15 cm apart, were fixed in the heel cups. FL was measured from the most posterior part of the calcaneus to the most distal part of the longest toe. TFL is defined as the foot length from the most posterior part of the calcaneus to the centre of the medial joint space of the first metatarsal phalangeal joint (30) Arch height index (AHI) was defined as the height of the arch at 50% total FL divided by TFL (32). Same measures were done with participants standing in order to acquire loaded foot dimensions (31). Also, the dorsal arch height difference between dorsal arch in a bipedal stance (AHIstand) and while sitting (AHIsit) was calculated, known as sit-to stand difference (32). Arch stiffness was calculated following Zifchock and colleagues' recommendations (32). A change in load between seating and standing conditions of 40% was assumed (value of change reflected the difference between half the body weight and the weight of the foot and shank) being the calculation as follows:

$$\text{Arch stiffness} = \frac{(0.40 \text{ x body mass})}{(\text{AHIsit} - \text{AHIstand})} \quad \text{(Equation 13)}$$

Every measure was repeated 3 times and the average was computed and used for analysis. The static foot posture and foot mobility measures have reported moderate to good intrarater

reliability (intraclass correlation coefficient = 0.81–0.99) and moderate to good interrater reliability (intraclass correlation coefficient = 0.58–0.99), respectively (31, 33).

Measurements of the running spatiotemporal parameters of CT (time the foot spends in contact with the ground) and FT (time from toes off to the initial contact of the same foot) (34) were done using the same photoelectric cell system (Microgate, Bolzano, Italy) described above, which was also previously validated for the assessment of gait spatiotemporal parameters of young adults (35). The system, calibrated by the manufacturer recommendations, was set on the treadmill surface for this study. The OptoGait system was linked to a laptop and the manufacturer software was used (Version 1.12.1.0, Microgate, Bolzano, Italy). Filter parameters GAitR-In and GAitR-Out were set at 0_0 (36). Furthermore, data was collected at a sampling frequency of 1,000 Hz and encrypted and stored carefully. Limb dominance was not taken into account (37).

Running gait stiffness reliability has been previously reported (38-40). Both Kvert (kN/m), defined as the ratio of maximal force to the vertical COM displacement at the middle of the stance phase, and Kleg (kN/m), defined as the ratio of the maximal force in the spring to the maximum leg compression at the middle of the stance phase (6), were calculated using the sine-wave method (8). For a runner's Kvert and Kleg estimation, Morin's approach requires us to collect a small amount of information (leg length, body mass, FT, CT, and speed). Actually, Morin (2005) showed that a small difference (0.67-6.93%) is found between stiffness when calculated using platform and the sine-wave methods (8). In addition, Pappas and colleagues (2014) endorsed that Morin's method determines accurately Kvert and Kleg for intra and inter-day designs (ICCs = 0.86-0.99) (41). It is found that Kvert presents lower coefficients of variation over a range of speeds than Kleg (38-40). It is worth mentioning that little difference has been reported in the reliability between measurements found using the sine-wave method (39, 40) and those using force platforms (38).

Statistical analysis

Descriptive data are presented as mean and standard deviation (SD). The normality distribution of the data was confirmed by Shapiro-Wilk's test ($p > 0.05$). In order to explore the between-sex differences, an analysis of variance (ANOVA) was conducted for stiffness parameters (i.e., variables considering body mass), and an analysis of covariance (ANCOVA), considering the body mass as covariate, was performed for jumping performance. The magnitude of the differences between values was also interpreted using the Cohen's d effect size (ES) (between-

group differences) (42). Effect sizes are reported as: trivial (<0.19), small (0.2-0.49), medium (0.5-0.79), and large (≥0.8) (42). Then, to analyse the relationship between parameters, a partial correlation analysis, adjusted by sex, was conducted. The following criteria were adopted to interpret the magnitude of correlations between measurement variables: <0.1 (trivial), 0.1–0.3 (small), 0.3–0.5 (moderate), 0.5–0.7 (large), 0.7–0.9 (very large) and 0.9–1.0 (almost perfect) (43). All statistical analyses were performed using SPSS software version 25.0 (SPSS Inc., Chicago, IL, USA) and statistical significance was accepted at an alpha level of 0.05.

RESULTS

A between-sex comparative analysis for stiffness-related parameters is provided in the Table 4. Significant differences between men and women were found in arch stiffness ($p = 0.026$, ES = 1.26) and Kvert_shod ($p = 0.025$, ES = 1.35), whereas no differences ($p \geq 0.05$, ES < 1.10) were found in the rest of parameters.

Table 4. Between-sex comparison in static (i.e., arch stiffness) and dynamic measures (i.e., vertical and leg stiffness) of lower-limb stiffness (±SD).

Variable	All (n=19)	Sex		p-value (ES)
		Men (n=14)	Women (n=5)	
Arch stiffness	733.23 (307.79)	824.13 (302.22)	478.70 (139.88)	0.026 (1.34)
Kvert_shod (kN/m)	23.32 (5.39)	24.92 (5.37)	18.83 (1.72)	0.025 (1.35)
Kleg_shod (kN/m)	8.10 (1.25)	8.42 (1.10)	7.21 (1.32)	0.062 (1.10)
Kvert_unshod (kN/m)	26.88 (3.13)	27.32 (3.19)	25.66 (2.91)	0.321 (0.61)
Kleg_unshod (kN/m)	11.69 (2.30)	11.39 (2.30)	12.53 (2.32)	0.356 (0.51)

ES: Cohen´s d effect size; Kvert: vertical stiffness; Kleg: leg stiffness; SD: standard deviation

Table 5 includes a sex comparison for DJ performance and RSI. Men showed a greater performance in DJ30 ($p = 0.038$, ES = 1.60) and higher values in both RSI20 ($p = 0.014$, ES = 1.67) and RSI30 ($p = 0.001$, ES = 1.78).

Table 5. Drop jump (DJ) performance parameters (±SD) and reactive strength index regarding sex

Variable	All (n=19)	Sex^		p-value (ES)
		Men (n=14)	Women (n=5)	
DJ20 (cm)	24.56 (6.86)	26.69 (6.30)	18.6 (4.75)	0.068 (1.42)
RSI20	2.17 (0.50)	2.35 (0.43)	1.66 (0.27)	0.014 (1.67)
DJ30 (cm)	27.48 (7.78)	30.05 (6.84)	20.20 (5.65)	0.038 (1.60)
RSI30	2.34 (0.52)	2.57 (0.39)	1.70 (0.87)	0.001 (1.78)

ES: Cohen´s d effect size; DJ20: jump height from a 20 cm drop jump; RSI20: reactive strength index calculated from DJ20; DJ30: jump height from a 30 cm drop jump; RSI30: reactive strength index calculated from DJ30; SD: standard deviation; ^ One-way analysis of covariance with body mass as covariates

A partial correlation analysis, adjusted by sex (Table 6), reported some significant relationships between lower-limb stiffness parameters (i.e., Kvert_shod and Kleg_shod: r =0.701, p < 0.01; Kleg_shod and Kvert_unshod: r = 0.600, p < 0.01; Kvert_unshod and Kleg_unshod, r = 0.738, p < 0.001) and RSI20 and RSI30 (r = 0.809, p < 0.001). No significant relationships were found between arch stiffness and the rest of parameters (r < 0.454, p ≥ 0.05). No significant relationships were found between reactivity and lower-limb stiffness (r < 0.245, p ≥ 0.05).

Table 6. Partial correlation analysis (r coefficient), adjusted by sex, between arch stiffness, lower-body stiffness during running in both shod and unshod conditions, and reactive strength index obtained from drop jumping.

	Arch stiffness	Kvert_shod	Kleg_shod	Kvert_unshod	Kleg_unshod	RSI20	RSI30
Arch stiffness	1	0.454	0.414	0.273	0.052	-0.062	-0.168
Kvert_shod		1	0.701**	0.333	-0.318	-0.011	-0.135
Kleg_shod			1	0.600**	0.249	0.101	-0.062
Kvert_unshod				1	0.738***	0.245	0.095
Kleg_unshod					1	0.209	0.169
RSI20						1	0.809***
RSI30							1

* p<0.05, ** p<0.01, *** p<0.001
Kvert: vertical stiffness; Kleg: leg stiffness; RSI20: reactive strength index calculated from a 20 cm drop jump; RSI30: reactive strength index calculated from a 30 cm drop jump

DISCUSSION

The present study sought to examine the relationship between reactivity and lower-limb stiffness in amateur endurance runners while jumping and running at 12 km·h^{-1}, as well as identifying possible sex differences. The major finding reported here was that no significant correlations were found between reactivity and lower-limb stiffness, suggesting that what may be reactive along the sagittal plane (i.e., DJ), may not be resembled over the horizontal plane (i.e., running). This statement seems to be supported by the specificity principle that claims that the task demands define the type of SSC used and, hence, the RSI values (15, 44), confirming, therefore, our hypothesis.

Running is an activity that involves eccentric-concentric muscle contraction. Stored elastic energy reutilization is considered a critical determinant of metabolic energy-saving mechanism during running. Reactive strength represents a runner's capacity to efficiently use the SSC and elastic energy produced by the musculotendinous unit (16). Several studies have used different jumping tests to identify the relationship between jump ability and distance running performance (45, 46). However, to the best of our knowledge, the correlation between RSI and

stiffness in different tasks remains uncertain. During running, muscles, tendons, and ligaments integrate as a spring to store and release elastic energy. The complex musculoskeletal system has been characterized using a spring-mass model (1). The findings reported here show sex differences for some parameters regarding lower-limb stiffness during running. Between men and women, very large correlations were found for both Kvert and Kleg at shod running, as well as for Kvert and Kleg at unshod running. Likewise, large correlations were found between Kleg while shod running and Kvert during unshod running.

Furthermore, our results show that the RSI reflects very large correlations with sex differences between RSI calculated from DJ20 and DJ30. For both men and women, the direction of the correlation was positive between RSI and DJ parameters, finding that the higher the drop, the greater the RSI value. These findings are opposed to those found by Kipp and colleagues, when they reported that DJ performance parameters such as RSI and DJ remained invariable across drop heights (20). The reason behind this discrepancy might be explained due to methodological differences between both studies. While Kipp and colleagues used 3 different heights (30, 45, and 60 cm) for analysis, our study considered only 2 different drop heights, 20 and 30 cm, since these two DJ heights have been suggested when assessing RSI (26, 47).

The RSI for females was significantly lower than for males in both DJ20 (1.66 ± 0.27 and 2.35 ± 0.43, respectively) and DJ30 (1.77 ± 0.87 and 2.57 ± 0.39, respectively). These values differ slightly from those found by Beattie and colleagues as they reported RSI values between 1.26 ± 0.24 and 1.50 ± 0.33 for DJ30. These fluctuations between outcomes in both studies might be explained based on different methods. While we split our sample into two groups in order to find sex-related differences, Beattie and colleagues do not specify whether their sample is made of only men, women or both (17). The work of Sole and colleagues support the present findings as they also reported greater RSI values for male (0.424 ± 0.108) than for female (0.314 ± 0.089) athletes (19). The large difference between our study and the Sole and colleagues' study for RSI values is explained by the different jumping tasks used. Sole and colleagues used CMJ, while in the current study DJ was in use following Beattie and colleagues' recommendations for RSI assessment during jumping (17). Assessment of Kvert during DJ correlates highly to RSI (48), meaning that RSI appears to reflect lower-limb stiffness in DJ, and, since stiffness-related sex differences have been thoroughly reported, it might also explain the differences in RSI values between men and women.

Men exhibited significant greater stiffness values than women in both static and dynamic conditions, particularly for arch stiffness and Kvert at shod running, what seems to agree previous sex-related stiffness studies (9, 10, 12). Similarly to previous work where males displayed greater Kvert (33.9 ± 8.7 kN/m) than females (26.3 ± 6.5 kN/m) (10), Kvert is significantly greater in males than in females (24.92 ± 5.37 and 18.83 ± 1.72 kN/m, respectively) in the present study. The disparity in values between both studies may be clarified due to dissimilar methods. While in our study participants were asked to run at 12 $km \cdot h^{-1}$ on a treadmill and stiffness was calculated by the sine-wave method (8), Granata and colleagues' participants were instructed to hop at their preferred frequencies on a force platform (10). The specificity of the different tasks, as well as the different methods to assess stiffness give enough evidence for the different values explanation between these two studies.

Additionally, particularly for women, it has been found that joint laxity is cyclic since when estrogen concentration increases over the menstrual cycle, knee laxity also increases (49). A 17% decrease in knee stiffness over the ovulatory phase resulting in a change in knee laxity from 13.35 ± 2.53 mm during the follicular phase to 14.43 ± 2.60 mm during ovulation (50). Since it is suggested that estrogen may increase collagen synthesis but decrease sinew stiffness (51), a reduction in tendon stiffness would be expected, affecting performance (52). Unfortunately, menstrual cycle was not considered in our study, limiting, therefore, the interpretation of our sex-related findings. Another limitation to be considered is that a motorised treadmill at 12 $km \cdot h^{-1}$ was used during the complete protocol. Furthermore, participants used their own running shoes during shod running trials, increasing, thus, the study ecological validity. Ultimately, the participants' foot-strike pattern was not considered.

This study determined the differences between RSI and sex-related correlations while jumping and lower-limb stiffness during running. Both the SSC and the lower-limb stiffness play an outstanding role within the neuromuscular behaviour while using elastic energy in sporting tasks such as running and jumping. Nevertheless, they may behave differently regarding the specificity principle behind each particular task. Both sport scientists and practitioners must consider sex-specific differences when assessing RSI and stiffness in female athletes as the musculotendinous properties vary across the menstrual cycle.

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DECLARATION OF INTEREST STATEMENT

The authors report no conflict of interest.

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Study 3. How do footwear, foot-strike pattern and step frequency influence on spatiotemporal parameters and lower-body stiffness in endurance running?

Under review in Journal of Biomechanics

ABSTRACT

The spring-mass model describes the running spring-like leg behaviour. However, the conditions in which running is performed influence the model's behaviour. This study aimed to determine the influence of footwear condition, foot-strike pattern, and step frequency on running spatiotemporal parameters and lower-body stiffness during treadmill running. Thirty-one amateur endurance runners performed a two-session protocol (shod and barefoot). Each session consisted of two trials at 12 $km \cdot h^{-1}$ over 5 minutes altering step frequency every minute (150, 160, 170, 180, 190 spm). First, participants were instructed to land first with the heel; after completion, the same protocol was repeated landing always first with the forefoot. Repeated measures ANOVAs resulted in significant differences for footwear condition, foot-strike pattern, and step frequency for each variable: percent contact time, percent flight time, vertical stiffness, and leg stiffness ($p < 0.001$). Interactions were found between foot-strike pattern and step frequency for spatiotemporal variables ($p = 0.027$) and between footwear condition and foot-strike pattern in the vertical ($p = 0.041$) and leg stiffness ($p = 0.044$) analysis. The results provide new insights on running biomechanics. Specifically, greater vertical and leg stiffness values were found when running barefoot. When forefoot running, higher stiffness values were observed. Likewise, both vertical and leg stiffness became greater as step frequency increased. Ultimately, the largest stiffness values were shown in the barefoot-forefoot condition. Lower-limb stiffness responds differently to changes in footwear condition, foot-strike pattern, and step frequency; thus, an appropriate manipulation might be beneficial when working on running retraining, performance, and injury prevention.

Keywords: *barefoot, foot-strike angle, long-distance running, spring-mass model, stiffness*

INTRODUCTION

The spring-mass model has been found to accurately describe the spring-like leg behaviour of running (1). This model includes a massless linear spring and a particle acting as the centre of mass of the whole body. The regulation of spatiotemporal and kinematic variables, vertical stiffness (Kvert) and leg stiffness (Kleg) are often used to characterise the spring-mass model features in individual runners. Whereas Kvert is the ratio of maximal force to maximal downward vertical displacement of the centre of mass, Kleg is defined as the ratio of the vertical ground reaction force to the leg-spring compression at the middle of the stance phase (2). Of particular importance to the aforementioned model are the conditions in which running is performed, which include footwear, foot-strike pattern, and step frequency.

In contrast to today's runners, who have the benefit of modern shoes with heels and cushions, most runners throughout history used footwear with minimal support (3). Trained subjects have previously been associated with significantly higher values for Kvert, Kleg, and step frequency while barefoot running as compared to shod runners (4). From a biomechanical perspective, the most reported difference between barefoot and shod running is the foot-strike pattern (3, 5, 6). Three primary foot-strike patterns for running have been established: a rear-foot strike (RFS), where the heel lands first; a mid-foot strike, heel and ball of the foot land simultaneously; and a forefoot strike (FFS), ball of the foot lands before the heel (3). Although barefoot runners tend to exhibit the FFS, most shod runners use the RFS as facilitated by the elevated and cushioned heel of the modern running shoes (3, 7).

The foot-strike pattern runners demonstrate is associated with the lower-limb stiffness' behaviour (3, 8, 9). Increased knee stiffness and decreased ankle stiffness is related to FFS, whereas for RFS the opposite has been reported (3, 8, 9). It has been reported that increase in ankle stiffness originated by a RFS pattern show higher influence on lower-limb stiffness, although this was tested during a hoping task (10). However, knee stiffness has been reported as the main influential factor on lower-limb stiffness while running using a FFS pattern (11). Running-related injuries are multifactorial. Endurance runners manage repeatedly the vertical ground reaction force impact, a collision force of about 1.5-3 times body weight, within the first 50 ms of the stance phase (3). Impacts associated with RFS running contribute to the high incidence of running-related injuries (3), especially tibial stress fractures and plantar fasciitis (12, 13).

Previous research suggests that either too much or too little stiffness throughout the lower extremity may induce musculoskeletal injuries (14, 15). Farley and González (2) showed that when humans increase their step frequency at a given running speed, the most important adjustment to the body's spring system is that leg spring becomes stiffer. As a result, the Kvert of the spring-mass system increases, the vertical displacement during the ground contact phase decreases, and the system bounces off the ground in less time. Likewise, Heiderscheit and colleagues (16) studied the effects of five different step frequencies on the hip, knee, and ankle joint energy absorption, but neither shod running nor foot-strike pattern were taken into account.

Ultimately, Almeida and colleagues (17) reported that the lack of standard methods comparing barefoot to shod running and regular to altered foot-strike patterns makes the analysis of the spring-mass model during running especially complex. As reported here, previous works have determined the influence of footwear, foot-strike pattern, and step frequency individually on lower-limb stiffness. Each of these variables influences the others and, therefore, lower-limb stiffness. To the authors' knowledge, there are no studies aimed at evaluating the behaviour of the spring- when footwear condition, foot-strike pattern, and step frequency are established as well as their relation to both Kvert and Kleg. Therefore, the current study sought to determine the influence of the footwear condition, foot-strike pattern, and step frequency on the running spatiotemporal parameters and lower-body stiffness during treadmill running. Based on previous studies, we hypothesised that the footwear condition, foot-strike pattern, and step frequency would contribute individually to lower-limb stiffness fluctuation. Likewise, we hypothesised that lower-limb stiffness would increase at barefoot running, using FFS, and as step frequency increase.

METHODS

Experimental Approach to the Problem

This study was conducted to identify the influence of footwear condition, foot-strike pattern, and step frequency on the spatiotemporal parameters and lower-body stiffness while running at a constant speed. The running spatiotemporal parameters of percentage of both ground contact time (%CT) and flight time (%FT) during the step cycle and Kvert and Kleg were analysed under twenty conditions: shod and barefoot running, two foot-strike patterns (FFS and RFS), and five step frequencies (150, 160, 170, 180, and 190 steps per minute). To clarify the influence of each variable and their interactions, a unilateral crossover design was

performed. This study was approved by the ethics committee at University San Jorge (Zaragoza, Spain).

Subjects

A sample of 31 habitually shod runners with a 10-Km-time range from thirty-nine to fifty-five minutes (18 males and 13 females; age: 34.42 ± 9.26 years; height: 171.54 ± 9.15 cm; body mass: 66.63 ± 11.3 kg; 10-Km time: 48.46 ± 3.85 min) volunteered in this study in March 2018. Participants were accustomed to treadmill running, not minimalist nor maximalist runners, and free from neuromuscular disorders and functional limitations for at least 6 months prior to participation. Informed consent, which complied with the standards of the Declaration of Helsinki of the World Medical Association (2013), was obtained. Participants ran with their own traditional running shoes (weight: 290.197 ± 32.826 gr; drop: 9.419 ± 1.928 mm; heel stack: 28.032 ± 3.351 mm). All participants were recruited amongst sport science students.

Procedures

This study was executed in the biomechanics laboratory of the University San Jorge across two different sessions separated by a one-week washout period. Participants ran shod during the first testing session and barefoot in the second session, and completed the entire protocol on a motorised treadmill with a maintained slope of 0% (HP cosmos Pulsar 4P; HP cosmos Sports & Medical, Gmbh, Nußdorf, Germany). Before both testing sessions, every subject performed a warm-up consisting of a 5-min continuous run and 5 min of active joint mobilization and dynamic stretching. None of the participants reported fatigue after the warm-up protocol. Since accommodation to running on a treadmill happens in 6-8 min for human locomotion (18, 19), an 8-minute accommodation program was executed by increasing speed by 1 $km \cdot h^{-1}$ every minute from 8 to 12 $km \cdot h^{-1}$ after warm up. Next, subjects ran at a speed of 12 $km \cdot h^{-1}$ for 5 minutes, altering step frequency every minute; the first 30 seconds of the minute were used to familiarise with the established step frequency (20) and the last 30 seconds were recorded for analysis. Recording 6–8 strides was adequate to acquire representative data for healthy adults (defined as 95% confidence intervals within 5% of error) (21). All participants began with a metronome-controlled (22) step frequency of 150 spm and were asked to land first on the heel (RFS). The protocol continued by modifying step frequency (160, 170, 180, and 190 spm). After that, participants performed the same step frequency ladder process however they were asked to land first on the forefoot (FFS). For the second session, participants returned and performed the same procedure under the barefoot condition.

The foot-strike pattern for each testing session was supervised by a trained researcher during the entire protocol and high-speed video was collected at 240 Hz (Imaging Source DFK 33UX174, The Imaging Source Europe GmbH; Germany). Range of interest (ROI) was adjusted to achieve 240 fps (1024x768 resolution). The camera was placed perpendicular to the treadmill from a sagittal view at 2 m from the centre of the treadmill and at a height of 0.30 m (23). Analysis was limited to measurements within ±2% of the established step frequency.

Materials and testing

For descriptive purposes, body height (cm) and body mass (kg) were determined using a precision stadiometer (SECA 222; SECA Corp., Hamburg, Germany) and a weighing scale (Tanita BC-601; TANITA Corporation, Maeno-Cho, Itabashi-ku, Tokyo, Japan). Participants wore only underwear during measurements.

The spatiotemporal parameters %CT and %FT were measured using OptoGait Photoelectric Cell system (Microgate, Bolzano, Italy), which was previously validated for the assessment of running gait spatiotemporal parameters of healthy adults (24). The system calibration was done by the manufacturer recommendations and consisted of two transmitting-receiving bars placed parallel to one another, and for this study, set on the treadmill surface (HP cosmos Pulsar 4P; HP cosmos Sports & Medical, Gmbh, Nußdorf, Germany). The OptoGait system was linked via a USB cable to a laptop and the manufacturer software was used (Version 1.12.1.0, Microgate, Bolzano, Italy). Filter parameters GAitR-In and GAitR-Out were set at 0_0 (25) and data was collected at a sampling frequency of 1,000 Hz and encrypted and stored securely. Limb dominance was not taken into account (26). Previous research (27) has described the spatiotemporal parameters analysed in this study:

- Contact time (CT, [s]): time one foot spends in contact with the ground on each step (i.e. from initial contact to the moment when the toes lifted off the ground).

- Flight time (FT, [s]): time from the toes lifting off to the initial contact of the next footfall.

- Step frequency (spm): number of ground contact events per minute.

- Percentage of ground CT (%CT) and FT (%FT) over the step cycle.

Kvert and Kleg were determined following Morin's sine-wave approach (28). Kvert (kN/m) is the ratio of maximal force to the vertical displacement of the centre of mass as its lowest point

(i.e., the middle of the stance phase) (2). Kleg (kN/m) is defined as the ratio of the maximal force in the spring to the maximum leg compression at the middle of the stance phase (2). Morin's method requires us to gather a relatively small amount of information (CT, FT, leg length, speed, and body mass) to estimate a runner's Kvert and Kleg. Indeed, Morin and colleagues demonstrated that there is a small difference (0.67-6.93%) between stiffness when calculated using the sine-wave and platform methods (28). Additionally, Pappas and colleagues ratified that the sine-wave approach can be used to measure Kvert and Kleg accurately for inter and intra-day designs with ICCs between 0.86-0.99 (29).

Statistical analysis

Descriptive statistics are represented as mean (±SD). Three-way within-subjects repeated measures ANOVAs ($\alpha = 0.05$) were run to compare the main effects of footwear (shod vs. barefoot), foot-strike pattern (FFS vs. RFS), and step frequency (150 - 190spm). Each dependent variable (%CT, %FT, Kvert, and Kleg) were assessed independently. For each of the ANOVA analyses, Mauchley's test of sphericity were violated, so the Greenhouse-Geisser correction factor was used in subsequent interpretation. For the step frequency condition, Bonferroni post-hoc pairwise comparisons were run when significant differences existed. Significant interactions of the three independent variables were assessed for each dependent variable. Partial eta squared ($\eta p2$) was calculated to provide estimates of effect size (30). Interpretations of small ($\eta p2 = 0.01$), medium ($\eta p2 = 0.06$), and large ($\eta p2 = 0.14$) were based on recommendations by Cohen (31). All statistical analyses were performed using SPSS (version 25, SPSS Inc., Chicago, IL, USA).

RESULTS

Footwear condition

The footwear condition comparison revealed significant differences in %CT (mean difference = -6.997 ± 0.882%; $p < 0.001$; $\eta p2 = 0.708$). %CT was significantly shorter under the barefoot condition compared to the shod condition for all step frequencies. Similarly, %FT was greater with the barefoot condition for all the step frequency (mean difference = 6.997 ± 0.882%; $p < 0.001$; $\eta p2 = 0.707$). The shod condition resulted in significantly lower Kvert and Kleg than the barefoot condition ($p < 0.001$; mean difference = 4.005 ± 0.686 and 4.038 ± 0.678 kN/m, respectively; $\eta p2 = 0.567$ and 0.577, respectively).

Foot-strike pattern

Greater %CT was found for RFS versus FFS pattern (mean difference = 7.451 ± 0.519%; $p < 0.001$; $\eta p2 = 0.888$). The mean %CT was 75.580 ± 0.932%, in comparison to 68.129 ± 0.943% in the FFS pattern. FFS pattern under the barefoot condition had the shortest %CT for all the step frequency studied (Figure 1). This is also supported in the main effects observed in %FT (FFS - RFS mean difference = -7.451 ± 0.519 %; $p < 0.001$; $\eta p2 = 0.888$). Kvert was greater under the FFS condition (mean difference = 1.731 kN/m; $p < 0.001$; $\eta p2 = 0.365$), and Kleg exhibited the same relationship (mean difference = 1.808 kN/m; $p < 0.001$; $\eta p2 = 0.360$).

Step frequency

Significant differences in step frequency conditions were observed for %CT and %FT ($p < 0.001$; $\eta p2 = 0.895$ and 0.895). Post-hoc analyses indicated that differences were significant across all conditions ($p < 0.001$). Generally, with each 10 spm increase in step frequency, the mean differences of %CT and %FT changed approximately +2.5% and -2.5%, respectively

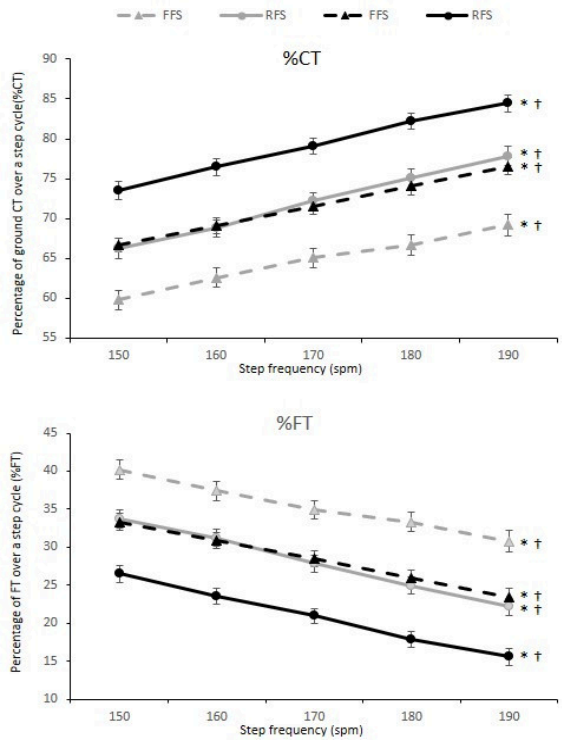

Figure 11. Mean values (±SD) of %CT (above) and %FT (below) during ladder step frequency protocol.

Grey lines represent barefoot running, black lines shod running.

* = $p < 0.001$ for significant differences between barefoot and shod conditions.

† = $p < 0.001$ for significant differences between FFS and RFS patterns.

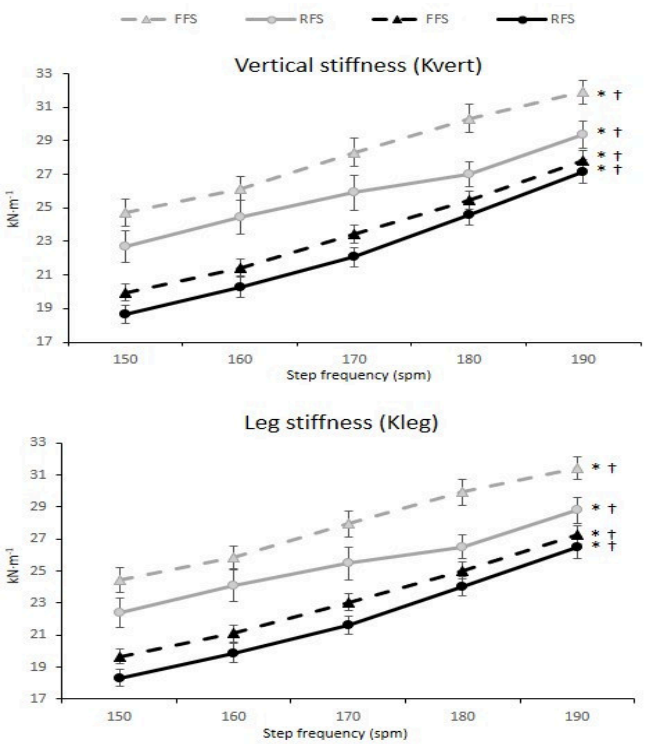

Figure 10. Mean values (±SD) of Kvert and Kleg during ladder step frequency protocol.

Grey lines represent barefoot running, black lines shod running.

* = $p < 0.001$ for significant differences between barefoot and shod conditions.

† = $p < 0.001$ for significant differences between FFS and RFS patterns.

(Figure 10). Additionally, significant main effects were observed for Kvert and Kleg ($p < 0.001$; $\eta p2 = 0.876$ and 0.873, respectively).

Post-hoc analyses indicated significant differences occurred between all step frequency measured for both variables ($p < 0.001$). As step frequency increased from 150 to 190 spm, Kvert and Kleg became significantly greater for all step frequencies (Figure 11).

Interactions

An interaction between foot-strike pattern and step frequency was observed for spatiotemporal variables of %CT ($p = 0.027$, $\eta p2 = 0.132$) and %FT ($p = 0.027$, $\eta p2 = 0.132$) as seen in Figure 11. It has been found that %CT increased and %FT decreased as step frequency became higher showing significant greater values at RFS running for both footwear conditions (barefoot and shod). The reverse is shown for %FT as it decreased alongside increased step frequency at RFS running (Figure 9). Because they are related variables, the significance ($p = 0.027$) effect size ($\eta p2 = 0.132$) were identical. An interaction between footwear condition and foot-strike pattern was observed in the Kvert and Kleg analyses ($p = 0.041$ and 0.044; $\eta p2 = 0.151$ and 0.147, respectively). When shod running, participants exhibited lower both Kvert and Kleg values, although these values were always greater at FFS running than at RFS running. The effect direction is shown in Figure 12. Although the described interactions were significant when evaluated at an alpha of 0.05, when familywise error rate was controlled for (6), these interactions became non-significant.

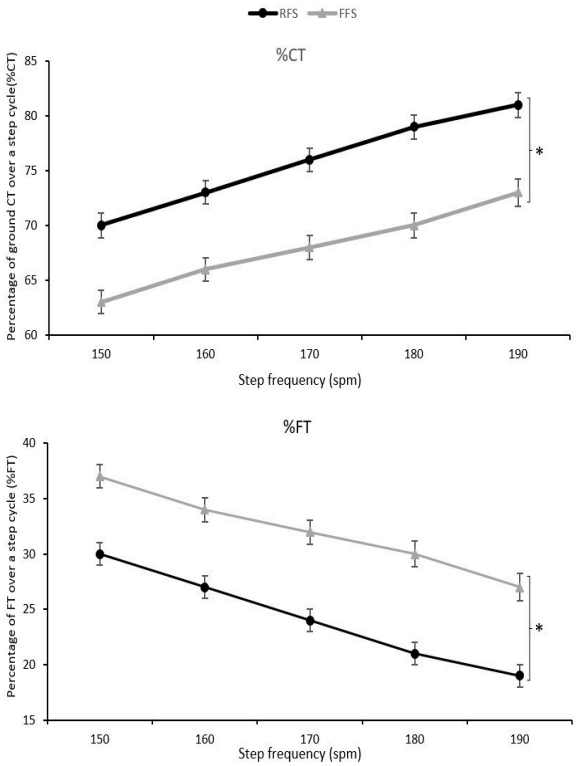

Figure 12. Effect direction of the interaction of foot-strike pattern-footwear conditions for the spatiotemporal parameters of %CT and %FT.
* = $p < 0.05$

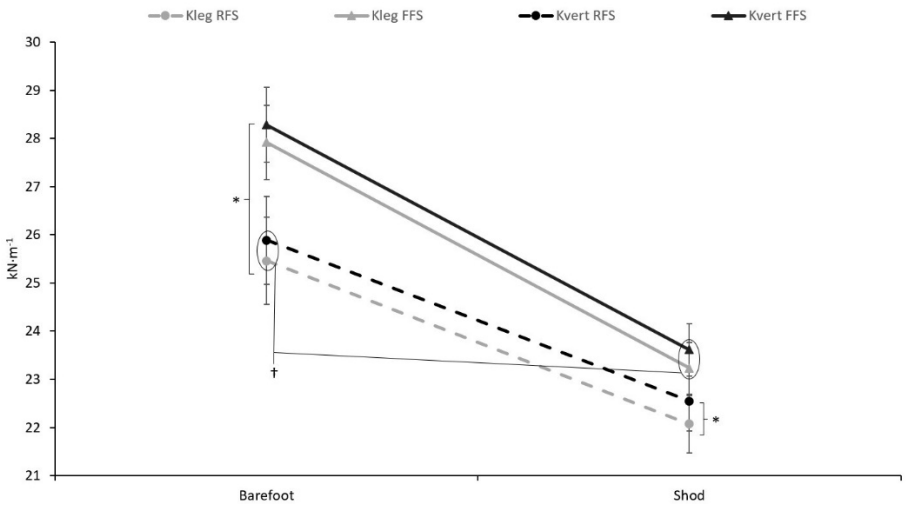

Figure 13. Effect direction of the interaction of foot-strike pattern-footwear conditions for lower-body stiffness. * = p < 0.001; † = p < 0.05.

DISCUSSION

This study sought to determine the influence of footwear condition, foot-strike pattern, and step frequency on spatiotemporal parameters and lower-body stiffness for healthy adults while treadmill running at a constant speed. Our results confirm our hypotheses showing that footwear condition (shod vs. barefoot), foot-strike pattern (RFS vs. FFS) and step frequency influence significantly %CT, %FT, Kvert and Kleg. Similarly, it has been confirmed that lower-limb stiffness exhibited the largest values when runners used a FFS pattern at barefoot running as step frequency increased.

The present study has reported that barefoot running decreases %CT and increases %FT in comparison to shod running. Contrary to Shih and colleagues (6), who claimed that shoe use may not influence spatiotemporal parameters, the present study shows significant differences between shod and barefoot conditions for %CT and %FT analysis. It is worth mentioning that while Shih et al. (6) did not control step frequency, this study considered its effects on the parameters analysed. Likewise, our findings suggested that foot-strike pattern is also associated with changes in spatiotemporal parameters. It has been demonstrated that whilst %CT values are lower for forefoot strikers than for rear-foot strikers, %FT values are higher. These values are justified by previous research (6). The current study also demonstrated that step frequency significantly influenced %CT and %FT; as step frequency increased from 150 to 190 spm, %CT significantly increased and %FT significantly decreased.

The non-significant interactions that were found between foot-strike pattern and step frequency in the analysis for the spatiotemporal variables likely do not jeopardize the main effects found

for foot-strike pattern or step frequency because of how large the effect sizes were. It has been demonstrated that the interaction of foot-strike pattern and step frequency occurs as step frequency increases. It is important to emphasise that although the percent of time spent in the stance phase increased, this was accompanied by a reduction in the absolute CT associated with increased step frequencies. This response may have occurred because the treadmill was run at a fixed speed and an increase in flight phase would accelerate the runner's speed.

Kvert and Kleg were higher for barefoot running, which aligns with previous studies (4, 6, 33). Higher values of Kvert and Kleg are shown for FFS (Kleg = 25.573 ± 0.495 kN/m) than RFS (Kleg = 23.765 ± 0.556 kN/m) showing similarities with the findings of Hamill and colleagues (8) (RFS Kleg = 15.97 ± 2.51 kN/m; FFS Kleg = 23.07 ± 5.52 kN/m). The difference in values between our study and Hamill's might be explained due to the different methods adopted; while we measured our participants running on a motorised treadmill by Morin's approach (28), Hamill's measures were done with a force platform flush with the ground where subjects ran across and performed ten running trials by stepping on the platform with the right foot without targeting the platform or altering stride features (8). Furthermore, in the current study, step frequency was controlled over the entire data collection, but Hamill (8) did not consider step frequency. The different foot-strike pattern might alter step frequency and, as shown in our data, cause a consequent alteration in Kvert and Kleg. Similarly, Kvert and Kleg were significantly affected by step frequency. As shown in our study, FFS running improves the leg's ability to store and release elastic energy by, as previously reported, increasing knee stiffness while decreasing ankle stiffness, as well as limiting knee range of motion (34). This may influence highly the overuse injury appearance since knee stiffness was found significantly higher in injured runners over a 2-year observational study (35). Our data proved that as step frequency increased from 150 to 190 spm, both Kvert and Kleg values became higher, which is consistent with an increase in the leg spring stiffness. This demonstrates that there were considerable adaptations to the behaviour of the musculoskeletal spring system with the alteration of step frequency as Farley and González described previously (2).

A non-significant interaction between footwear conditions and foot-strike pattern was also found for the Kleg and Kvert outcome variables. This interaction suggests the reduction in stiffness that occurs with the addition of the shoe is greater with a FFS than a RFS. Ultimately, the Kvert and Kleg reported from the barefoot+RFS condition is highly reduced compared to the barefoot+FFS condition, potentially suggesting that participants alter mechanical strategy to attenuate the shock occurred with a RFS. This suggestion can be supported by Lieberman et

al. (3), who suggested that ground collision forces are less with barefoot+FFS running. As runners move from shod to barefoot running some alterations happen such as their tendency to adopt a flatter strike when landing affecting, amongst others, step frequency, loading rate, Kvert, and Kleg (3, 36). Of note, the present study found that the spatiotemporal parameters of %CT and %FT were almost identical for barefoot+RFS and shod+FFS conditions suggesting that the advantages derived from adopting a FFS pattern when shod running might be almost equivalent to the effect produced by barefoot running using a RFS pattern. A potential mechanism of participant's reduction in Kvert and Kleg might be the vertical gain during the flight phase (thus reducing max force and attenuating shock), however this hypothesis cannot be supported from the current study.

There are some limitations to be considered. It is well known that the sine-wave approach (28) calculates lower-limb stiffness indirectly reporting a small difference (0.67-6.93%) in comparison with stiffness measured using the platform method. For Kvert and Kleg, the differences found here for footwear, foot-strike pattern, and step frequency reported error values above the upper limit described in such method for all the conditions (6.9-16.3%). Although footwear while shod running trials was not standardised, all runners wore their own traditional running shoes; thus, the ecological validity of the study was increased. Furthermore, the participants were habitually shod runners, then, the novelty of the task might influence our outcomes at barefoot running. The distinction between habitually RFS and FFS runners was not considered for the study, showing, therefore, a lack of intergroup interpretation. Ultimately, the entire protocol was developed on a motorised treadmill at a constant velocity and only with amateur runners, remaining unknown the likely outcomes over ground and with non-amateur runners.

CONCLUSION

This study showed that Kvert and Kleg increase when a forefoot strike pattern is adopted, when barefoot running, and as step frequency becomes higher while treadmill running. Furthermore, when assessing lower-limb stiffness, foot-strike pattern, footwear condition, and step frequency need to be considered as each of these factors influences the others. The findings reported provide insights on the biomechanical behaviour of the spring-mass model under different conditions and their proper quantification and manipulation may facilitate our understanding of running performance, injury prevention, and training.

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Table 7. Summary of main results obtained in the current PhD Thesis

Study	Results
1. Test–retest reliability of the OptoGait system for the analysis of spatiotemporal running gait parameters and lower body stiffness in healthy adults	Although mean comparisons revealed significant differences between measurements in spatiotemporal running gait characteristics and lower body stiffness for intra-session ($p < 0.05$ in all parameters), the effect size was always small (<0.4). Moreover, the relationship between measurements was very large for spatiotemporal parameters and lower body stiffness ($r > 0.7$). The intraclass correlation coefficients revealed an almost perfect correlation between measurements (intraclass correlation coefficients >0.81), except Kleg with substantial reliability (intraclass correlation coefficient = 0.788). The inter-session reliability revealed some significant differences in contact time ($p = 0.009$) and Kleg ($p = 0.013$), although Cohen's d indicated small effect size (<0.31). The relationship between sessions was very large for spatiotemporal parameters and lower body stiffness ($r > 0.8$), and the intraclass correlation coefficients revealed an almost perfect inter-session association (intraclass correlation coefficients >0.881). The results found here show that the OptoGait system can be used confidently for running spatiotemporal parameters analysis and lower body stiffness at a constant velocity for healthy adults.
2. Is there a relationship between reactivity and stiffness in amateur endurance runners? A comparative analysis between sexes	Repeated measures ANOVAs resulted in significant sex differences for arch stiffness and vertical stiffness at shod running, showing men greater values ($p < 0.05$). ANCOVA resulted in significant sex differences for reactivity strength index and (men showed greater values for both heights) and 30-cm drop jump performance ($p < 0.05$). Cohen's d was used to interpret effect size. A sex-adjusted partial correlation showed no significant relationships between reactivity and lower-limb stiffness ($r < 0.245$, $p > 0.05$). The results indicate that the spring-mass model reacts differently to tasks based on their specificity principle. Additionally, sex-related differences must be considered when assessing the stretch-shortening cycle.
3. How do footwear, foot-strike pattern and step frequency influence on spatiotemporal parameters and lower-body stiffness in endurance running?	Repeated measures ANOVAs resulted in significant differences for footwear condition, foot-strike pattern, and step frequency for each variable: percent contact time, percent flight time, vertical stiffness, and leg stiffness ($p < 0.001$). Significant interactions were observed between foot-strike pattern and step frequency for spatiotemporal variables ($p = 0.027$) and between footwear condition and foot-strike pattern in the vertical ($p = 0.041$) and leg stiffness ($p = 0.044$) analysis. The results indicate that lower-limb stiffness responds differently to changes in footwear condition, foot-strike pattern, and step frequency.

6. DISCUSSION

Over the course of this work, the main influencing factors on lower-limb stiffness were identified and discuss from both injury prevention and performance perspective. Furthermore, it has been shown that the OptoGait system can be used with confidence for running spatiotemporal parameters analysis and lower body stiffness at a constant velocity. Moreover, the findings demonstrate that spring-mass model behaviour varies according to the task's specificity principle. Besides, sex differences must be taken into account when evaluating the stretch-shortening cycle. Finally, lower-limb stiffness responds differently to changes in influencing factors such as footwear condition, foot-strike patter, and step frequency.

Over the course of this research, the test-retest reliability (i.e., intra- and inter-session) of the OptoGait system for the acquisition of spatiotemporal gait characteristics and lower-body stiffness while running on a treadmill was analysed. The study tested 31 healthy adults to demonstrate the reliability of the OptoGait system, while acquiring the running spatiotemporal parameters of FT, CT, SL, and SF, as well as both Kvert and Kleg. The results indicate that spatiotemporal parameters and lower-body stiffness during running were reliable in both intra- and inter-session contexts. Nevertheless, the Bland-Altman analysis provides insights into the systematic differences between the measurements. None of the measured variables reported heteroscedasticity of error, except vertical and leg stiffness in the session 1 vs. session 2 comparison (i.e., inter-session reliability). The results reinforce the intra- and inter-session reliability data for spatiotemporal parameters and intra-session reliability for lower-body stiffness, but also highlight the lack of stability of the Kvert and Kleg variance.

Reliability is essential for a running gait analysis system to guarantee that differences in running gait performance are related to gait changes as opposed to errors in data collection. The findings reported in this PhD Thesis are in line with previously reported results regarding the spatiotemporal parameters for healthy adults [1,2]. Gomez Bernal et al. assessed the reliability of the OptoGait system for spatiotemporal parameters analysis while walking on a treadmill. Similarly, Lee et al. asked their participants to walk three times on a walkway at a comfortable velocity. One of the studies executed over this PhD Thesis shows the test-retest reliability of the OptoGait system for treadmill running spatiotemporal parameters analysis. Compared to previous studies where running spatiotemporal parameters were measured using the OptoGait system [3], an incremental velocity protocol (10 to 20 $km \cdot h^{-1}$) was implemented in such prior studies to measure running spatiotemporal parameters in contrast to the study here described,

where a constant velocity (12 km·h^{-1}) was established during data collection to examine the test-retest reliability of the OptoGait system for treadmill running spatiotemporal parameters. Due to the lack of available information regarding the use of the OptoGait system for spatiotemporal parameters while running at a constant velocity, it makes comparison to other studies difficult, thus underscoring the importance of this study.

It has been demonstrated that the value of Kvert is always higher than Kleg in locomotion since variations in leg length surpass those of the centre of mass [4,5]. Despite Kvert and Kleg being derived from analogous mechanical concepts, they are not equivalent and adapt differently to fluctuations in running conditions [4,6]. Hence, the evaluation of both Kvert and Kleg is justified. The findings reported here of the intra-session trials correlate perfectly with those found by Pappas et al. [5] as shown respectively in the following parentheses regarding ICCs for FT (0.904 and 0.970), SL (0.948 and 0.925), SF (0.943 and 0.932), and Kvert (0.956 and 0.972) and differ slightly for CT (0.865 and 0.986) and Kleg (0.788 and 0.982). In regard to the findings of the inter-session trials, the results found in the current study are very close to Pappas et al.'s results as shown respectively in the following parentheses regarding ICCs for CT (0.900 and 0.925), FT (0.894 and 0.902), SL (0.916 and 0.860), SF (0.921 and 0.863), Kvert (0.896 and 0.922), and Kleg (0.881 and 0.873). The slight differences between both studies could be explained due to difference in methods. While Pappas et al. only included male participants, both male and female runners were included for the current study. Moreover, Pappas and colleagues recorded three rounds of 30 seconds at 16 km·h^{-1} for each participant compared to the current study where data for each participant was recorded once over three minutes at a constant velocity of 12 km·h^{-1}. It has been demonstrated that longer recording periods return smaller systematic bias and random errors, as well as narrower limits of agreement regarding step variability [7].

While developing the present PhD Thesis, the relationship between reactivity and lower-limb stiffness in amateur endurance runners while jumping and running at 12 km·h^{-1}, as well as identifying possible sex differences was also clarified. The major finding reported here was that no significant correlation was found between reactivity and lower-limb stiffness, suggesting that what may be reactive along the sagittal plane (i.e., DJ), may not be reflected over the horizontal plane (i.e., running) in amateur endurance runners. This statement seems to be supported by the specificity principle that states that the task demands define the type of SSC used and, therefore, the RSI values [3,8], consequently confirming our hypothesis.

Running is an activity in which eccentric-concentric muscle contraction is involved. Stored elastic energy reutilization is considered a critical determinant of metabolic energy-saving mechanism during running. Reactive strength represents a runner's capacity to efficiently use the SSC and elastic energy produced by the musculotendinous unit [9]. Several studies have used different jumping tests to identify the relationship between jump ability and distance running performance [10,11]. However, the correlation between RSI and stiffness in different tasks remains uncertain. During running, muscles, tendons, and ligaments integrate as a spring to store and release elastic energy. The complex musculoskeletal system has been described based on the spring-mass model [12]. The findings derived from this PhD Thesis demonstrate that lower-limb stiffness during running correlated to sex differences for some parameters. Between men and women, very large correlations were found for both Kvert and Kleg when shod running, as well as for Kvert and Kleg at unshod running. Likewise, large correlations were found between Kleg while shod running and Kvert during unshod running.

Furthermore, these results show that the RSI reflects very large correlations with sex differences between RSI calculated from DJ20 and DJ30. For both men and women, the direction of the correlation was positive between RSI and DJ parameters, showing that the higher the drop, the greater the RSI value. These findings are opposed to those found by Kipp and colleagues, when they reported that DJ performance parameters such as RSI and DJ remained invariable across drop heights [13]. The reason behind this discrepancy may be credited to methodological differences between both studies. While Kipp and colleagues used 3 different heights (30, 45, and 60 cm) for analysis, our study considered only 2 different drop heights, 20 and 30 cm, since these two DJ heights have been suggested when assessing RSI [14].

The RSI for females was significantly lower than for males in both DJ20 (1.66 ± 0.27 and 2.35 ± 0.43, respectively) and DJ30 (1.77 ± 0.87 and 2.57 ± 0.39, respectively). These values differ slightly from those found by Beattie and colleagues where RSI values between 1.26 ± 0.24 and 1.50 ± 0.33 for DJ30 were reported. Again, methodological differences may account for these discrepancies. While for this particular study the sample was split into two groups in order to find sex-related differences, Beattie and colleagues do not specify whether their sample consisted of only men, women or both [9]. Sole and colleagues endorsed the present study findings as they also reported greater RSI values for male (0.424 ± 0.108) than for female (0.314 ± 0.089) athletes [15]. The large difference between our study and the Sole's study for RSI values is explained by the different jumping tasks used. Sole and colleagues used CMJ, while in the current study DJ was in use following Beattie and colleagues' recommendations for RSI

assessment during jumping [9]. Assessment of Kvert during DJ correlates highly to RSI [16], meaning that RSI appears to reflect lower-limb stiffness in DJ, and, since stiffness-related sex differences have been thoroughly reported, it could be attributable to the differences in RSI values between men and women.

Men exhibited significant greater stiffness values than women in both static and dynamic conditions, particularly for arch stiffness and Kvert at shod running, which seems to confirm previous sex-related stiffness studies [17-19]. Similarly to previous work where males displayed greater Kvert (33.9 ± 8.7 kN/m) than females (26.3 ± 6.5 kN/m) (17), Kvert is significantly higher in males than in females (24.92 ± 5.37 and 18.83 ± 1.72 kN/m, respectively). The disparity in values between both studies may be clarified due to dissimilar methods. While in this PhD study participants were asked to run at 12 $km \cdot h^{-1}$ on a treadmill and stiffness was calculated by the sine-wave method [4], Granata and colleagues' participants were instructed to hop at their preferred frequencies on a force platform [17]. The specificity of the different tasks, as well as the different methods to assess stiffness, provide enough evidence to explain the different values between these two studies.

The final aim of this research sought to determine the influence of footwear condition, foot-strike pattern, and step frequency on spatiotemporal parameters and lower-body stiffness for healthy adults while treadmill running at a constant speed. The results derived from this particular study confirm that footwear (shod vs. barefoot), foot-strike pattern (RFS vs. FFS) and step frequency influence significantly %CT, %FT, Kvert and Kleg, as stated in the hypotheses of the current PhD Thesis. Similarly, it has been confirmed that lower-limb stiffness exhibited the largest values when runners used a FFS pattern at barefoot running as step frequency increased.

It has been reported that barefoot running decreases %CT and increases %FT in comparison to shod running. Contrary to Shih and colleagues [20], who argued that shoe use may not influence spatiotemporal parameters, while pursuing this PhD Thesis significant differences between shod and barefoot conditions for %CT and %FT analysis have been shown. It is worth mentioning that while Shih et al. [20] did not control step frequency, one of the PhD studies described here considered its effects on the parameters analysed. Likewise, the findings reported suggest that foot-strike pattern is also associated with changes in spatiotemporal parameters. It has been demonstrated that whilst %CT values are lower for forefoot strikers than for rear-foot strikers, %FT values are higher. These values are justified by previous

research [20]. It has been also shown that step frequency significantly influenced %CT and %FT; as step frequency increased from 150 to 190 spm, %CT significantly increased and %FT significantly decreased.

The non-significant interactions that were found between foot-strike pattern and step frequency in the analysis for the spatiotemporal variables likely do not jeopardize the main effects found for foot-strike pattern or step frequency because of how large the effect sizes were. It has been demonstrated that the interaction of foot-strike pattern and step frequency occurs as step frequency increases. It is important to emphasise that although the percent of time spent in the stance phase increased, this was accompanied by a reduction in the absolute CT associated with increased step frequencies. This response may have occurred because the treadmill was run at a fixed speed and an increase in flight phase would accelerate the runner's speed.

Kvert and Kleg were higher for barefoot running, which aligns with previous studies [20-22]. Higher values of Kvert and Kleg are shown for FFS (Kleg = 25.573 ± 0.495 kN/m) than RFS (Kleg = 23.765 ± 0.556 kN/m) showing similarities with the findings of Hamill and colleagues (14) (RFS Kleg = 15.97 ± 2.51 kN/m; FFS Kleg = 23.07 ± 5.52 kN/m). The difference in values between the implemented PhD study and Hamill's might be explained due to the different methods adopted; while in this study our participants were measured running on a motorised treadmill by Morin's approach [4], Hamill's measures were done with a force platform flush with the ground where subjects ran across and performed ten running trials by stepping on the platform with the right foot without targeting the platform or altering stride features [23]. Furthermore, in the PhD study, step frequency was controlled over the entire data collection, but Hamill [23] did not consider step frequency. The different foot-strike pattern might alter step frequency and, as shown in the data of the mentioned PhD study, cause a consequent alteration in Kvert and Kleg. Similarly, Kvert and Kleg were significantly affected by step frequency. As shown in this study, FFS running improves the leg's ability to store and release elastic energy by increasing knee stiffness while decreasing ankle stiffness, as well as limiting knee range of motion [24]. This may influence highly the overuse injury appearance since knee stiffness was found significantly higher in injured runners over a 2-year observational study [25]. The data here reported proved that as step frequency increased from 150 to 190 spm, both Kvert and Kleg values became higher, which is consistent with an increase in the leg spring stiffness. This demonstrates that there were considerable adaptations to the behaviour of the musculoskeletal spring system with the alteration of step frequency as Farley and González described previously [6].

A non-significant interaction between footwear conditions and foot-strike pattern was also found for the Kleg and Kvert outcome variables. The interaction suggests the reduction in stiffness that occurs with the addition of the shoe is greater with a FFS than a RFS. Ultimately, the Kvert and Kleg reported from the barefoot+RFS condition is highly reduced compared to the barefoot+FFS condition, potentially suggesting that participants alter mechanical strategy to attenuate the shock occurred with a RFS. This suggestion can be supported by Lieberman et al. [26], who suggested that ground collision forces are less with barefoot+FFS running. As runners move from shod to barefoot running some alterations happen such as their tendency to adopt a flatter strike when landing affecting, amongst others, step frequency, loading rate, Kvert, and Kleg [26,27]. Of note, in the last of the PhD studies was found that the spatiotemporal parameters of %CT and %FT were almost identical for barefoot+RFS and shod+FFS conditions suggesting that the advantages derived from adopting a FFS pattern when shod running might be almost equivalent to the effect produced by barefoot running using a RFS pattern. A potential mechanism of participant's reduction in Kvert and Kleg might be the vertical gain during the flight phase (thus reducing max force and attenuating shock), however this hypothesis cannot be supported from the current study.

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7. LIMITATIONS

Over this section, the different limitations found while developing the different studies building this PhD Thesis are described.

All the studies here presented have been done over a motorised treadmill. Although this might be a limitation, it has recently been demonstrated that spatiotemporal parameters, kinetics and kinematics, muscle activity, and musculotendinous outcome measures are to a large extent comparable between over ground and treadmill running [1]. Nevertheless, the authors of the systematic review mentioned above claimed that kinematic differences at FSP when inferring treadmill running biomechanics to over ground running should be considered [1]. Moreover, the velocity for all the studies was established at 12 $km \cdot h^{-1}$. It is worth noting that over ground running allows the ability to alter velocity but by establishing a constant velocity one of the influential factors on the spring-mass model and spatiotemporal parameters is kept under control, facilitating the subsequent analysis of lower-limb stiffness during running.

Another limitation to be considered might be the use of Morin's sine-wave method to measure Kvert and Kleg [2]. Although it is not a direct method, Morin's approach shows good accuracy and efficacy for the analysis of lower-body stiffness (a difference between 0.67-6.93% compared with stiffness measured with force platforms) as described in the introduction section. In consequence, one of the studies while developing this PhD Thesis proved that for Kvert and Kleg, the differences found for footwear, foot-strike pattern, and step frequency reported error values above the upper limit described in such method for all the conditions (6.9-16.3%). Likewise, as the population assessed while pursuing this PhD Thesis has entirely been amateur-runner healthy adults, it is still unclear the likely outcomes for different ages, athletic levels and pathological population.

Particularly in the Study 2, when assessing sex differences, menstrual cycle was not considered in the study, limiting, therefore, the interpretation of our sex-related findings. It is suggested that estrogen might increase the synthesis of collagen but decreases sinew stiffness [3], thus, a tendon stiffness reduction is expected affecting, consequently, performance [4].

It is also worth mentioning that during the research done to accomplish Study 3, all the participants ran habitually shod and the novelty of running without shoes could have an effect on the outcomes at barefoot running. Additionally, differentiation amongst FFS and RFS

runners was not taken into consideration, thus, there was no possibility to show intergroup interpretation.

Ultimately, all the participants used their own traditional running shoes increasing, therefore, the ecological validity of all the studies increased.

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8. FUTURE PERSPECTIVE

Knowledge about lower-limb stiffness in running and both its modulation and alteration as a tool for improvement and retraining play an outstanding role in endurance training plans. A deeper knowledge about the behaviour of the different musculoskeletal structures influencing the spring-mass model and the SSC would provide all the sport science community members some insight on how to vary different factors according to the desired outcome. However, as discussed over this PhD Thesis, many variables influence the behaviour of lower-limb stiffness, thus, further research needs to be done.

This PhD Thesis examined the reliability of the OptoGait photoelectric cell system for spatiotemporal parameters and lower-body stiffness while running for healthy adults, all of them amateur runners. In order to broad the scope of action of the OptoGait system, due to its friendly use, further studies analysing the test-retest reliability of this system for both well-trained endurance as well as pathological runners are needed.

The role of the foot in the modelling of the spring-mass behaviour needs deeper research. As the foot is the ultimate structure of the lower body in contact with the floor in sporting tasks such as running and jumping, it is worth noting that further research is needed in order to understand how foot structure influence on the behaviour of all the structures which take part in the spring-mass model (i.e., muscle and tendon activity or joint stiffness). Likewise, it would be of a great value to understand how lower-limb tendons (i.e., Achilles and patellar tendons) work regarding the foot-strike pattern adopted by a runner.

Likewise, further research in the analysis of sex differences regarding the behaviour of both the spring-mass model and the SSC is needed. Menstrual cycle and its effects on musculotendinous tissues must be considered in future work since injury risk and performance are altered.

Ultimately, additional research is needed under the scope of running-related injuries. Overuse injuries are predominant amongst runners and the development of protocols aiming at evaluating the injury risk of a runner might be crucial for all the running community, from the very low level to the highest competitive one.

9. CONCLUSION

I. A list of influencing factors on lower-limb stiffness and their relation to both performance and injuries while running has been supplied. The information shown in this work contributes to further solidify, increase, and broaden the knowledge of the behaviour of lower-body stiffness and its scope of action. It has been demonstrated that these factors influence one another, thus, they should not be measured individually.

II. The OptoGait system performs reliable evaluation for running spatiotemporal parameters analysis and lower-body stiffness at a constant velocity for healthy adults. The findings reported here might have a high value for sport scientists and clinicians working on both running gait retraining and improvement. The user-friendliness of the OptoGait system and its proved reliability for running spatiotemporal parameters and lower-limb stiffness analysis provide coaches and clinicians a trustworthy instrument to make judgements regarding the degree of change related to the normal variability of measuring between trials or sessions by contributing, especially, to the early identification of running pathologies.

III. Differences between RSI and sex-related correlations while jumping and lower-limb stiffness during running were found. Both the SSC and the lower-limb stiffness play an outstanding role within the neuromuscular behaviour while using elastic energy in sporting tasks such as running and jumping. Nevertheless, they may behave differently regarding the specificity principle behind each particular task. Both sport scientists and practitioners must consider sex-specific differences when assessing RSI and stiffness in female athletes as the musculotendinous properties vary across the menstrual cycle.

IV. It has been shown that Kvert and Kleg react differently to alterations in footwear condition, FSP, and SF while treadmill running. It has been shown that one variable influences the others, thus, the behaviour of the spring-mass model should be analysed cautiously when altering factors such as footwear, FSP or SF; the presence or absence of running shoes might contribute to the alteration of the FSP a runner adopts, what influences SF and, therefore, lower-limb stiffness at a given velocity. The findings reported provide insights on the biomechanical performance of the spring-mass model under different conditions and their proper quantification

and manipulation may facilitate our understanding of injury management, training, and racing in running

10. APPENDIX

10.1. Appendix 1 – Study 1 PDF-Proof for publication

Original Article

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Test–retest reliability of the OptoGait system for the analysis of spatiotemporal running gait parameters and lower body stiffness in healthy adults

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Abstract

Despite the widespread use of the OptoGait photoelectric cell system for the analysis of running spatiotemporal parameters, its reliability has not been proved. Consequently, this study intends to determine the test–retest reliability of the system when applied to treadmill running spatiotemporal parameters and lower body stiffness at a constant velocity. Amateur endurance runners (n = 31; age: 34.42 ± 9.26 years; height: 171.54 ± 9.15 cm; body mass: 66.63 ± 11.3 kg) voluntarily consented to participate in this study. Data for each participant were recorded twice per session across two testing sessions. The intra-session and inter-session reliabilities of spatiotemporal parameters and lower body stiffness were determined through test–retest analysis. Although mean comparisons revealed significant differences between measurements in spatiotemporal running gait characteristics and lower body stiffness for intra-session ($p < 0.05$ in all parameters), the effect size was always small (< 0.4). Moreover, the relationship between measurements was very large for spatiotemporal parameters and lower body stiffness ($r > 0.7$). The intraclass correlation coefficients revealed an almost perfect correlation between measurements (intraclass correlation coefficients > 0.81), except Kleg with substantial reliability (intraclass correlation coefficient = 0.788). The inter-session reliability revealed some significant differences in contact time ($p = 0.009$) and Kleg ($p = 0.013$), although Cohen's d indicated small effect size (< 0.31). The relationship between sessions was very large for spatiotemporal parameters and lower body stiffness ($r > 0.8$), and the intraclass correlation coefficients revealed an almost perfect inter-session association (intraclass correlation coefficients > 0.881). The results found here show that the OptoGait system can be used confidently for running spatiotemporal parameters analysis and lower body stiffness at a constant velocity for healthy adults.

Keywords

Reliability, testing, OptoGait, running, stiffness

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Introduction

Running is an enduringly popular pursuit. Benefits include improved cardiovascular function and mental health, stress relief, and enjoyment.[1–4] When animals run, they bounce along the ground. Such movement is facilitated by a system of musculoskeletal springs, comprised of muscles, tendons and ligaments which store elastic energy when stretched and release it when recoiled.[5–7] During running, this complex musculoskeletal system behaves much like a single linear spring (the 'leg spring').[8] In fact, a simple spring-mass model consisting of a single linear leg spring and a mass equivalent to that of the animal has been shown to describe

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and predict the mechanics of running remarkably well.[9–14] As vertical stiffness (Kvert) and leg stiffness (Kleg) influence the regulation of both spatiotemporal and kinematic variables, they are usually used while identifying these characteristics in individual runners. The Kvert (kN/m) is the ratio of maximal force to the vertical displacement of the centre of mass as its lowest point is reached (i.e. the middle of the stance phase). Similarly, Kleg (kN/m) is defined as the ratio of the maximal force in the spring to the maximum leg compression at the middle of the stance phase.[8,15]

While some previous studies described the influence of contact time (CT) and Kleg on both performance and running economy,[16,17] others have not demonstrated this influence.[18,19] Limitations of the methods in use for running biomechanics analysis might be the main reason for this difference. The drawbacks to commercially available tools for such analysis include limited accessibility, high cost, sensory fragility, and operating complexity, and they are mainly employed in research rather than clinical settings. It has been shown that high-speed video analysis is a reliable and valid method to measure running kinematics,[20] as well as three-dimensional (3D) motion capture system – considered as a 'gold standard'. However, running kinematics analysis using the systems mentioned above requires, among others, highly trained users for proper data collection, as well as data analysis. Floor-level, high-density photoelectric cells (OptoGait; Microgate, Bolzano, Italy), which are portable and allow quantification of spatiotemporal gait parameters on most flat surfaces, are used for clinical purposes.[21]

Although previous research concerning the OptoGait™ system has considered its reliability in assessing spatiotemporal walking and racewalking gait variables,[21–23] measuring spatiotemporal gait characteristics during running by implementing an incremental speed protocol[24,25] and calculating both Kvert and Kleg while running on a treadmill with different slope gradients,[26] the system reliability for the analysis of running gait spatiotemporal parameters, as well as lower body stiffness, is still unknown. Therefore, the aim of this study is to analyse the test–retest reliability of spatiotemporal gait characteristics and lower body stiffness while running on a treadmill at a constant velocity by comparing data intra-session and inter-sessions.

Methods

An observational study, aligned with STROBE guidelines,[27] was conducted for accuracy diagnosis of running gait. The running spatiotemporal parameters of CT, flight time (FT), step length (SL), and step frequency (SF) were analysed, including both Kvert and Kleg. This study was approved by the ethics committee at the University San Jorge (Zaragoza, Spain).

Participants

A sample of 31 healthy subjects, 18 men and 13 women (age: 34.42 ± 9.26 years; height: 171.54 ± 9.15 cm; body mass: 66.63 ± 11.3 kg), who were accustomed to running on a treadmill and able to run 10 km in 50–60 min, voluntarily participated in this study. Informed consent, which complied with the standards of the Declaration of Helsinki of the World Medical Association, was obtained from all participants prior to the study. Subjects who reported musculoskeletal injuries sustained within the previous 6 months or suffered from any other impairment that might affect their running gait were excluded from the study. Consequently, participants were free from cardiovascular, neurologic, or musculoskeletal conditions and familiar with running on a treadmill. The recruitment was done among sport sciences students.

Procedures

This study was executed in the biomechanics laboratory at the University San Jorge across two different sessions. Participants performed the same protocol under the same conditions. They were instructed by a researcher and completed the entire protocol running on a treadmill with an established inclination of 0%. Subjects started warming up at a speed of 8 km/h, increasing it freely over the course of 8 min ultimately reaching 12 km/h because previous studies[28,29] have shown that accommodation to treadmill running on human locomotion takes approximately 6–8 min. After the warm-up, participants ran at a speed of 12 km/h for 3 min during which time data were recorded for analysis. Subsequently, subjects ran for 5 min at a self-selected speed. Then, they ran again for three more minutes at 12 km/h with data being recorded for analysis. Subjects left the biomechanics laboratory after completing the running. One week later, subjects returned and repeated the same procedure under the same conditions. Subjects were instructed to continue their regular training, but were asked to avoid competitions and high-intensity activities 24 h for the study. All the steps occurred in the sensor area during analysis. Besser et al.[30] showed that recording 6–8 strides was adequate to acquire representative data for healthy adults (defined as 95% confidence intervals within 5% of error).

Both body mass (kg) and height (cm) for each participant were found using a weighting scale (Tanita BC-601; TANITA Corporation, Maeno-Cho, Itabashi-ku, Tokyo, Japan) and a precision stadiometer (SECA 222; SECA Corp., Hamburg, Germany), respectively. Participants wore only underwear during these measures. The leg length (L) of each participant was found in accordance with Winter's[31] anthropometric equations as shown in equation (1)

$$L = 0.53h \tag{1}$$

where h stands for the participant's height (m).

The running spatiotemporal parameters of CT (s), FT (s), SL (cm), and SF (spm) were measured using the

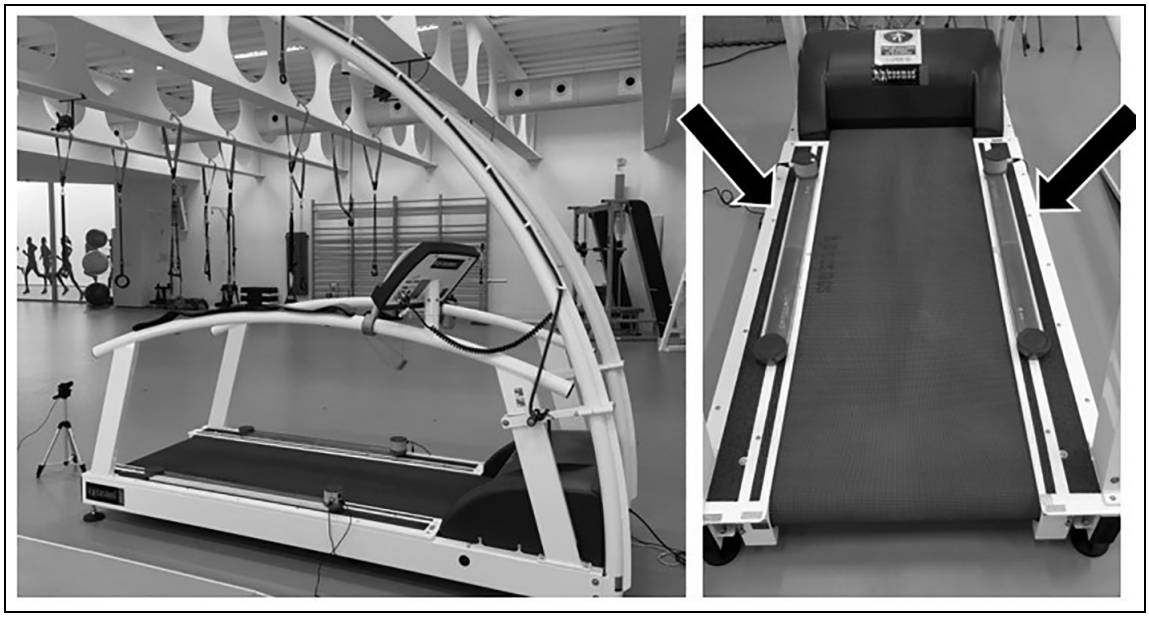

Figure 1. Location of the OptoGait system on a treadmill.

OptoGait Photoelectric Cell system (OptoGait) – previously validated for the evaluation of spatiotemporal features of the gait in young adults.[23] The OptoGait system calibration was done by the manufacturer and consisted of two transmitting–receiving bars placed parallel to one another, set on the treadmill surface for this study (HP cosmos Pulsar 4P; HP cosmos Sports & Medical, Gmbh, Nußdorf, Germany; Figure 1). The OptoGait system was linked via a USB cable to a personal computer and the manufacturer's software was used (Version 1.12.1.0) to minimise the systematic bias, the filter parameters GAitR-In and GAitR-Out were both set at 0_0.[23,32] The data were extracted at a sampling frequency of 1000 Hz, encrypted and stored on a computer. According to Brown et al.,[33] limb dominance was not considered.

This study employs a procedure developed by Morin et al.[15] to determine lower body stiffness. Kvert (kN/m) determines the general level of stiffness in the body by finding the ground reaction force and vertical displacement of the centre of mass relationship, whereas Kleg (kN/m) shows the stiffness in just the lower part of the body (feet, ankles, and hip joints) and gives the ratio between the ground reaction force and the deformation in leg length.[15] Morin's approach is useful because it only requires gathering a relatively small amount of information (CT, FT, leg length, speed, and body mass) to calculate the runner's approximate Kvert and Kleg. Indeed, these authors have demonstrated that there is a small difference (0.67%–6.93%) between stiffness when calculated using the sine-wave and platform methods.[15] For their part, Pappas et al. confirmed that the sine-wave method could be used to accurately measure Kvert and Kleg for intra- and inter-day designs with ICCs between 0.86 and 0.99.[34]

Data analysis

Descriptive statistics are represented as mean (±SD). Tests of normal distribution and homogeneity by the Kolmogorov–Smirnov and Levene's test, respectively, were conducted on all data before analysis. A mean comparison analysis (T-test) was conducted between variables from both measurements (i.e. intra-session) and from both days (i.e. inter-session). The magnitude of the differences was interpreted using Cohen's d effect size (ES).[10] ESs are reported as: trivial (< 0.2), small (0.2–0.49), medium (0.5–0.79), and large ($\geqslant 0.8$).[10] The relationship and association of variables from different measurements (i.e. intra-session) and from different testing days (i.e. inter-session) were quantified through the Pearson correlation coefficient (r) and the intraclass correlation coefficient (ICC). The following criteria were adopted to interpret the magnitude of correlations between measurement variables: < 0.1 (trivial), 0.1–0.29 (small), 0.3–0.49 (moderate), 0.5–0.69 (large), 0.7–0.89 (very large), and 0.9–1.0 (almost perfect).[11] Based on the characteristics of this experimental design and following the guidelines reported by Koo and Li,[35] the authors decided to conduct a 'two-way random-effects' model (ICC [2,k]), 'mean of measurements' type, and 'absolute' definition for the ICC measurement. The interpretation of the ICC was based on the benchmarks reported by a previous study:[13] ICC < 0 reflects 'poor', 0–0.20 'slight', 0.21–0.40 'fair', 0.41–0.60 'moderate', 0.61–0.80 'substantial', and > 0.81 'almost perfect' reliability. The Bland–Altman[14] limits of agreement method (mean difference ± 1.96 SD) was used to analyse differences in spatiotemporal features and lower body stiffness between measurements (i.e. intra-session) and between testing sessions (i.e. inter-session). Heteroscedasticity of error was defined as an $r^2 > 0.1$.

Table 1. Intra-session reliability of spatiotemporal parameters and lower limb stiffness.

Variable	Measurement 1 (± SD)	Measurement 2 (± SD)	p-value (Cohen's d)	Pearson coefficient (r)	ICC (95% CI)
CT (s)	0.274 (0.020)	0.280 (0.017)	0.006 (0.323)	0.811***	0.865 (0.670–0.940)
FT (s)	0.085 (0.021)	0.081 (0.019)	0.055 (0.199)	0.841***	0.904 (0.795–0.954)
SL (cm)	120.40 (5.48)	121.45 (5.08)	0.012 (0.209)	0.919***	0.948 (0.874–0.977)
SF (spm)	166.55 (7.60)	164.91 (7.30)	0.007 (0.214)	0.911***	0.943 (0.854–0.975)
Kvert (kN/m)	22.19 (3.41)	21.71 (3.26)	0.049 (0.148)	0.924***	0.956 (0.905–0.979)
Kleg (kN/m)	7.33 (1.01)	6.96 (0.89)	0.009 (0.389)	0.700***	0.788 (0.521–0.902)

SD: standard deviation; ICC: intraclass correlation coefficient; CI: confidence interval; CT: contact time; FT: flight time; SL: step length; SF: step frequency; Kvert: vertical stiffness; Kleg: leg stiffness.
***$p < 0.05$.

Table 2. Inter-session reliability of spatiotemporal parameters and lower limb stiffness.

Variable	Day 1 (±SD)	Day 2 (±SD)	p-value (Cohen's d)	Pearson coefficient (r)	ICC (95% CI)
CT (s)	0.274 (0.020)	0.268 (0.018)	0.009 (0.315)	0.850***	0.900 (0.743–0.957)
FT (s)	0.085 (0.021)	0.089 (0.022)	0.446 (0.186)	0.814***	0.894 (0.771–0.951)
SL (cm)	120.40 (5.48)	120.59 (5.57)	0.649 (0.07)	0.843***	0.916 (0.819–0.961)
SF (spm)	166.55 (7.60)	166.12 (7.85)	0.799 (0.052)	0.852***	0.921 (0.828–0.963)
Kvert (kN/m)	22.19 (3.41)	22.40 (4.23)	0.064 (0.05)	0.896***	0.896 (0.770–0.952)
Kleg (kN/m)	7.33 (1.01)	7.60 (1.25)	0.013 (0.238)	0.833***	0.881 (0.709–0.948)

SD: standard deviation; ICC: intraclass correlation coefficient; CI: confidence interval; CT: contact time; FT: flight time; SL: step length; SF: step frequency; Kvert: vertical stiffness; Kleg: leg stiffness.
***$p < 0.05$.

All the statistical analyses have been executed following the suggestions done by Atkinson and Nevill for assessing reliability.[36] The level of significance used was $p < 0.05$. Data analysis was performed using the SPSS (version 21; SPSS Inc., Chicago, IL, USA).

Results

The intra-session reliability of spatiotemporal parameters and lower body stiffness were determined through test–retest analysis (Table 1). Despite mean comparisons (i.e. measurement 1 vs measurement 2) which revealed significant differences between measurements in spatiotemporal gait characteristics and lower body stiffness ($p < 0.05$ in all parameters), the ES was always small (< 0.4). In addition, the relationship between measurements was very large for spatiotemporal parameters and lower body stiffness ($r > 0.7$). The ICCs also revealed an almost perfect association between measurements (ICCs > 0.81), except from Kleg with substantial reliability (ICC = 0.788).

The pairwise comparisons between testing days (i.e. inter-session reliability) revealed some significant differences in CT ($p = 0.009$) and Kleg ($p = 0.013$), although Cohen's d indicated small ES (< 0.31; Table 2). The relationship between sessions was very large for spatiotemporal parameters and lower body stiffness ($r > 0.8$), and the ICCs revealed an almost perfect inter-session association (ICCs > 0.881).

Through Bland–Altman plots, Figures 2 and 3 show the intra-session differences between the measurements (systematic bias and random error) and the degree of agreement (95% limits of agreement). Small biases and errors were observed in spatiotemporal parameters (CT: -0.01 ± 0.01 s; FT: 0.00 ± 0.01 s; SL: -1.04 ± 2.16 cm; SF: 1.63 ± 3.15 spm; Figure 1) and vertical and leg stiffness (Kvert: 0.48 ± 1.31 kN/m; Kleg: 0.37 ± 0.74 kN/m; Figure 2). No heteroscedasticity of error was found in any variable ($r^2 < 0.1$).

Bland–Altman plots also show the inter-sessions differences in the measured variables and the degree of agreement between the two measurements. Small systematic biases and random errors were reported for spatiotemporal parameters (CT: 0.01 ± 0.01 s; FT: 0.00 ± 0.01 s; SL: 0.27 ± 3.05 cm; SF: -0.20 ± 4.15 spm) with no heteroscedasticity of error ($r^2 < 0.1$; Figure 4). As for vertical and leg stiffness, despite biases and errors, they were small (Kvert: -0.78 ± 2.12 kN/m; Kleg: -0.35 ± 0.69 kN/m), while heteroscedasticity of error was found in both variables (Kvert: $r^2 = 0.472$; Kleg: $r^2 = 0.107$; Figure 5).

Discussion

This study aimed to analyse the test–retest reliability (i.e. intra- and inter-session) of the OptoGait system for the acquisition of spatiotemporal gait characteristics and lower body stiffness while running on a treadmill. The study tested 31 healthy adults to demonstrate

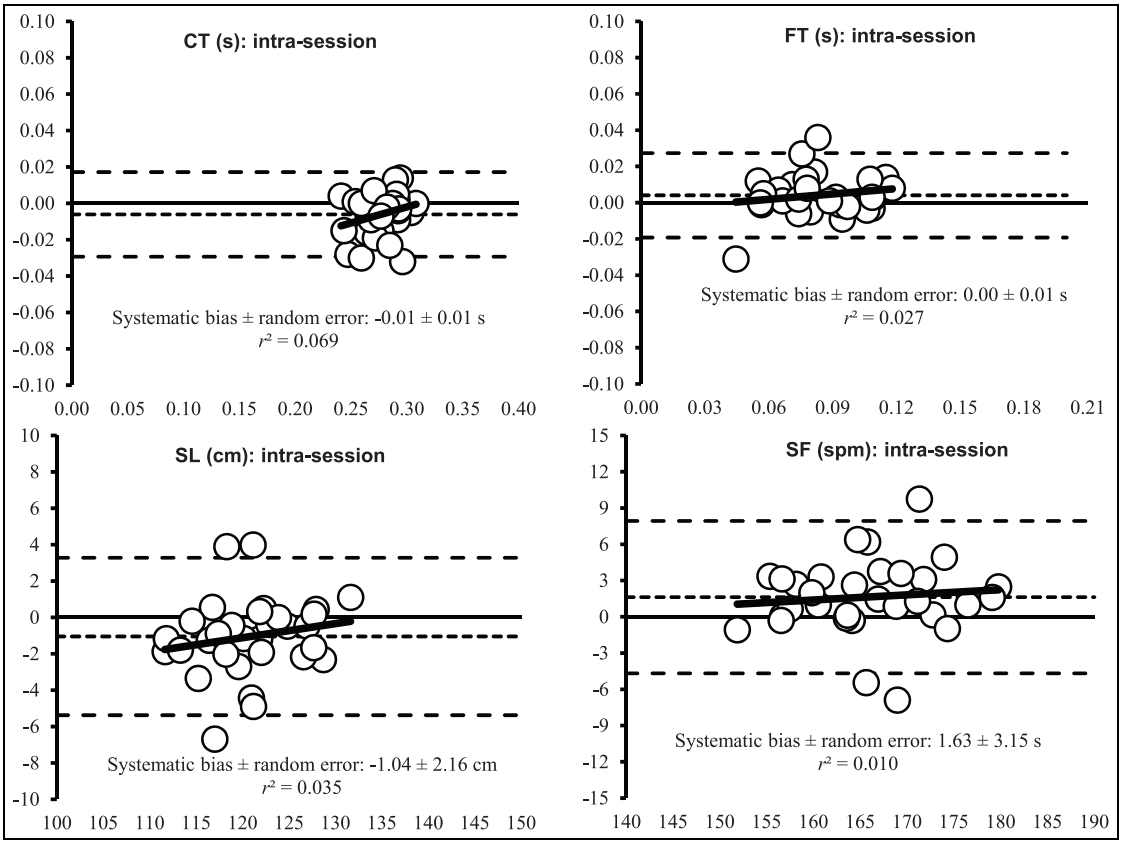

Figure 2. Intra-session differences between the measurements (systematic bias and random error) and the degree of agreement (95% limits of agreement) for CT, FT, SL, and SF.

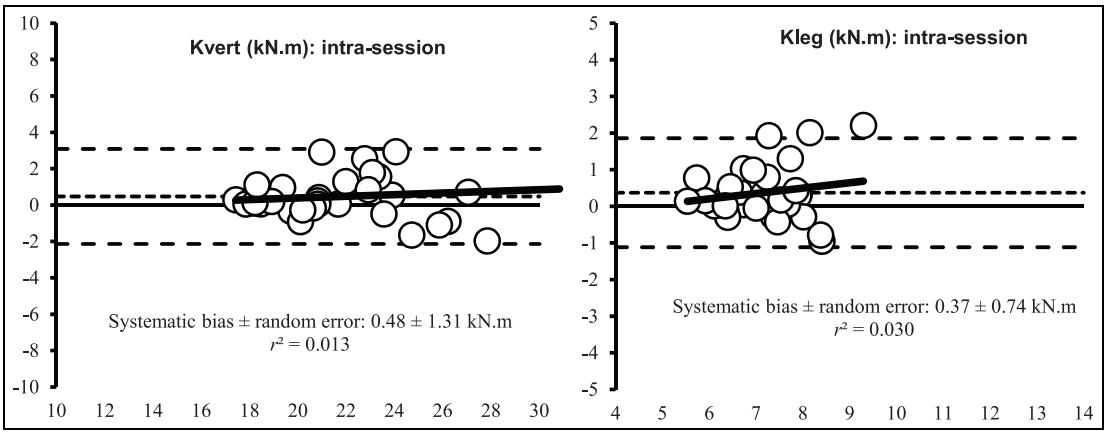

Figure 3. Intra-session differences between the measurements (systematic bias and random error) and the degree of agreement (95% limits of agreement) for Kvert and Kleg.

the reliability of the OptoGait system, while acquiring the running spatiotemporal parameters of FT, CT, SL, and SF, as well as both Kvert and Kleg. The results indicate that spatiotemporal parameters and lower body stiffness during running were reliable in both intra- and inter-session contexts. Nevertheless, the Bland–Altman analysis provides insights into the systematic differences between the measurements. None of the measured variables reported heteroscedasticity of error, except vertical and leg stiffness in the session 1 versus session 2 comparison (i.e. inter-session reliability). The results reinforce the intra- and inter-session reliability data for spatiotemporal parameters and intra-session reliability for lower body stiffness, but it also warns about the lack of stability of the Kvert and Kleg variance.

Reliability is essential for a running gait analysis system to guarantee that differences in running gait

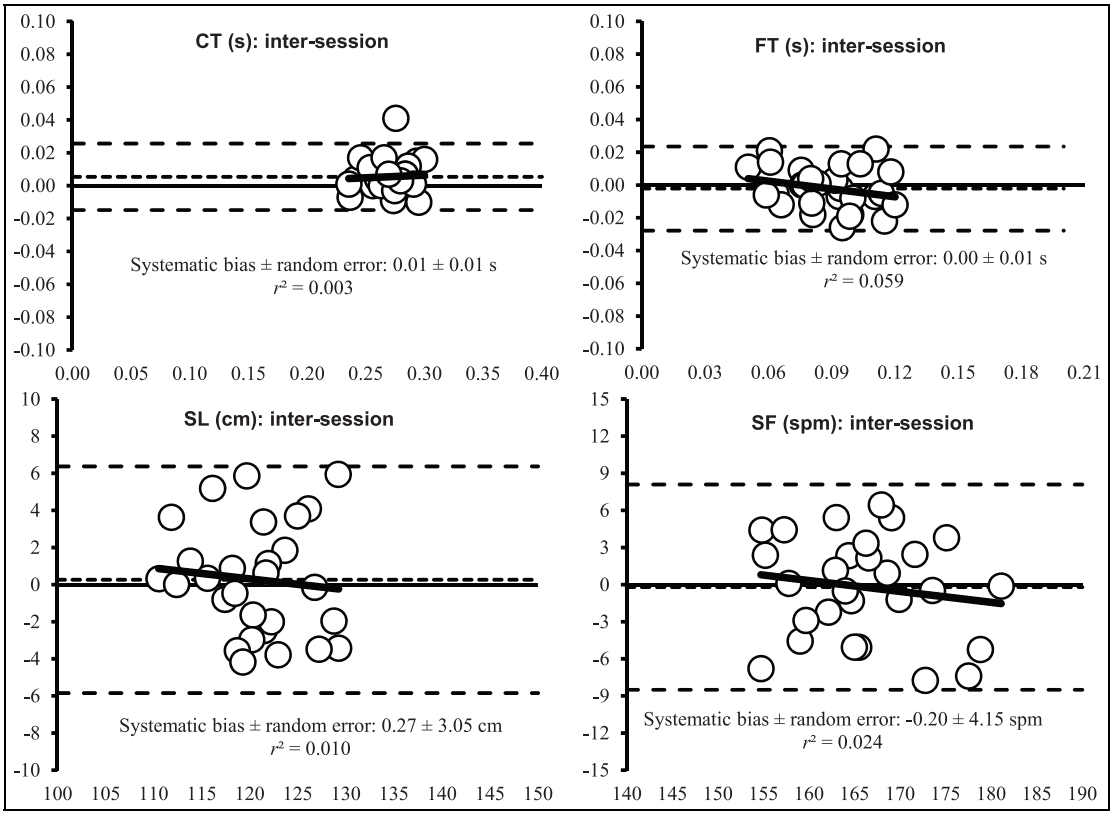

Figure 4. Inter-session differences between measurements and degree of agreement for CT, FT, SL, and SF.

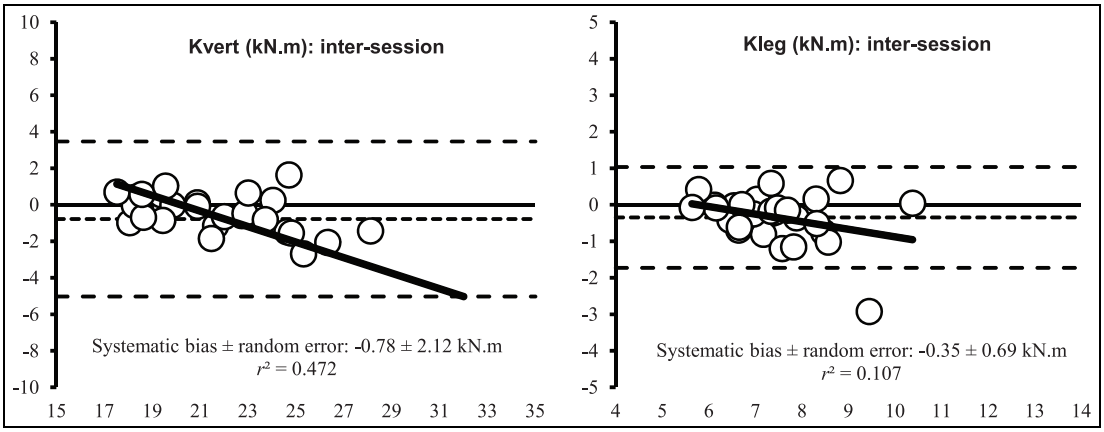

Figure 5. Inter-session differences between measurements and degree of agreement for Kvert and Kleg.

performance are related to gait changes as opposed to errors in data collection. The current findings are similar to previously reported results regarding the spatiotemporal parameters for healthy adults.[21,22] While Gomez Bernal et al. tested the reliability of the OptoGait system for spatiotemporal parameters analysis while walking on a treadmill and Lee et al. asked their participants to walk three times on a walkway at a comfortable velocity, this study shows the test–retest reliability of the OptoGait system for treadmill running spatiotemporal parameters analysis. Compared with previous studies where running spatiotemporal parameters were measured using the OptoGait system,[25] an incremental velocity protocol (10–20 km/h) was implemented in various studies to measure running spatiotemporal parameters in contrast to this study, where a constant velocity (12 km/h) was established during data collection to examine the test–retest reliability of the OptoGait system for treadmill running spatiotemporal parameters. Due to the lack of available information regarding the use of the OptoGait system for spatiotemporal parameters while running at a constant velocity, it makes comparison to other studies difficult, thus underscoring the importance of this study.

It has been demonstrated that the value of Kvert is always higher than Kleg in locomotion since variations in leg length surpass those of the centre of mass.[15,34] Despite Kvert and Kleg being derived from analogous mechanical concepts, they are not equivalent and adapt differently to fluctuations in running conditions.[8,15] Hence, the evaluation of both Kvert and Kleg is justified. The findings reported here of the intra-session trials correlate perfectly with those found by Pappas et al.[34] as shown respectively in the following parentheses regarding ICCs for FT (0.904 and 0.970), SL (0.948 and 0.925), SF (0.943 and 0.932), and Kvert (0.956 and 0.972) and differ slightly for CT (0.865 and 0.986) and Kleg (0.788 and 0.982). In regard to the findings of the inter-session trials, the results found in this study are very similar to Pappas et al'.s results as shown respectively in the following parentheses regarding ICCs for CT (0.900 and 0.925), FT (0.894 and 0.902), SL (0.916 and 0.860), SF (0.921 and 0.863), Kvert (0.896 and 0.922), and Kleg (0.881 and 0.873). The slight differences between both studies might be related to differences in methods. While Pappas et al. only included male participants, the participants for this study included both male and female runners. Moreover, Pappas and colleagues recorded three rounds of 30-s at 16 km/h for each participant compared with this study where data for each participant were recorded once over 3 min at a constant velocity of 12 km/h. It has been demonstrated that longer recording periods return smaller systematic bias and random errors, as well as narrower limits of agreement regarding step variability.[37]

Although this study sheds some light on the use of the OptoGait system as a reliable tool for the analysis of running spatiotemporal parameters, some limitations must be considered. On one hand, the laboratory scene should be considered while interpreting these findings; nevertheless, participants were accustomed to running on a treadmill. On the other hand, although Morin et al'.s[15] approach shows good efficacy and accuracy for the analysis of lower body stiffness, it is not a direct method. The strong reliability of the OptoGait system demonstrated by the current results will provide future researchers enough evidence to use this photoelectric system for the accuracy analysis of running spatiotemporal parameters and lower body stiffness. Since healthy adults have been evaluated in this study, future research work should consider the assessment of the system for different ages and population suffering from musculoskeletal pathologies.

Conclusion

This study shows that the OptoGait system performs reliable evaluation for running spatiotemporal parameters analysis and lower body stiffness at a constant velocity for healthy adults. The findings reported here might have a high importance for sport scientists and clinicians working on both running gait retraining and improvement. The user-friendliness of the OptoGait system and its proved reliability for running spatiotemporal parameters analysis provide coaches and clinicians a trustworthy instrument to make judgements regarding the degree of change related to the normal variability of measuring between trials or sessions, especially for early identification of running pathologies.

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Declaration of conflicting interests

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10.2. Appendix 2 – FI-379 Study 1

Written acceptance of the co-authors of a research publication for its presentation as part of a PhD Thesis

Aceptación escrita de los coautores para que el doctorando presente el trabajo como tesis doctoral

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ACEPTACIÓN ESCRITA DE LOS COAUTORES PARA QUE EL DOCTORANDO PRESENTE EL TRABAJO COMO TESIS DOCTORAL / ***Written acceptance of the co-authors of a research publication for its presentation as part of a PhD Thesis***

Datos del coautor / *co-author data*

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10.3. Appendix 3 – Stiffness in Running: A narrative integrative review

Stiffness in Running: A narrative integrative review

Under review in Strength and Conditioning Journal

ABSTRACT

Although the study of running spatiotemporal parameters has contributed to obtaining a deeper knowledge about the spring-mass model and its capacity to estimate and predict kinetic and kinematic variables, the contribution of lower-limb stiffness to this model is not fully explored. While its impact on athletic performance seems considerable, recognition of lower-limb stiffness among coaches and practitioners remains sparse. This review is aimed at detecting influencing factors on lower-limb stiffness during running and to discussing these factors from an injury prevention and performance perspective. The findings reported integrate our current knowledge of lower-body stiffness during running and offer new scopes of scientific attention. It is strongly recommended not to measure the effect of different variables on lower-limb stiffness individually as they influence each another. The spring-mass model behaviour when altering variables such as footwear or foot-strike pattern needs cautiously examination. Although both stretch-shortening cycle (SSC) muscle power and stiffness are key parameters when elastic strain energy is stored and reutilised in given sports and exercise disciplines, hormonal fluctuations (i.e., caused by the menstrual cycle) should draw special attention in female athletes since affecting musculotendinous stiffness properties and thereby influencing athletic performance and injury prevalence.

Keywords: injury; leg stiffness; performance; running; stretch-shortening cycle

INTRODUCTION

The increasing population participating in running events at all ages and levels has increased the interest for running research in the scientific community. Consequently, research activities analysing the health benefits derived from running have focused on various physiological as well as social and psychological aspects. Others have intended to clarify the mechanisms of running-related injuries, and many works aimed at determining physiological and biomechanical parameters of importance for running performance.

Spring-mass model

During the running stance phase, the leg resembles the behaviour of a spring which compresses and decompresses in a cyclic manner (16), with gravitational and inertial forces exerted on the body mass (m) representing the main source of leg-spring compression (156) (Figure 1). The mechanics of horizontal running can be predicted extraordinarily well using the linear leg spring model with a mass equivalent to that of the runner (2, 7, 8, 16, 100). During the leg-spring compression phase, represented by the eccentric phase of stance, mechanical energy is stored. This stored energy is released as elastic energy recoil in the subsequent concentric phase where muscle-tendon forces are declining (156). Contact phases in which the system rotates forward over a monopodial support alternate with float phases where the system behaves ballistically (147). Initial angle of attack (Ɵ) and the stiffness of the leg-spring play key roles in the concluding stage of the spring-mass model (156).

The stretch-shortening cycle (SSC) (146) and lower-limb stiffness (109), particularly vertical stiffness (Kvert) and leg stiffness (Kleg), are the two most important neuromuscular elements linked to elastic energy utilization during running. Therefore, the influencing factors on lower-limb stiffness for both running performance and running-related injuries based on the spring-mass approach are analysed in this narrative review by focusing on integrative experimental reports in the literature.

STIFFNESS

The term stiffness originated in classical physics; specifically as a component of Hooke's Law. This law is defined as $F=kx$, where F signifies the force required to deform (stretch or compress) an object, k is the stiffness constant, and x is the distance the object is deformed (14). The stiffness constant, k, is also known as the spring constant. It represents the stiffness of an ideal spring mass system (14), since the above formula can be rearranged into $k=\Delta F/\Delta x$, where ΔF and Δx are change in force and length respectively (90). Thus, expressed at its simplest, stiffness describes the relationship between a given force and the magnitude of deformation of an object or body (12, 14).

Stiffness is a frequently employed concept in relation to characterising human movement or describing neuromuscular function (14, 90, 119, 130). Where the physical body is concerned, the term can be applied across a wide breadth of levels, from that of a single muscle fibre to modelling the entire body as spring-mass system (12, 14). Given its elastic properties, the leg-spring tends to resist to any deforming force. The magnitude of such resistance is dependent upon the stiffness of the leg-spring. Similarly, during the phase of leg-spring decompression, stiffness positively correlates with the magnitude of returned elastic energy (27). Increased musculotendinous unit stiffness would be expected to maximise energy conversion from potential, stored within the elastic components of the lower limb during eccentric lengthening, to kinetic, released during the phase of subsequent contractile shortening (54).

Investigation into the relationship between stiffness and athletic performance sees four measurements commonly defined (13, 98):

1. *Vertical stiffness* ($kN \cdot m^{-1}$) describes the global compression of a runner (86), that is, vertical movement of their centre of mass expressed in relation to the concurrent

change in vertical GRF (ΔGRF / Δdisplacement) typically assessed in the sagittal plane (90).

2. *Leg stiffness* ($kN \cdot m^{-1}$) refers to how the various elements of the leg spring (i.e. muscles, tendons, ligaments) behave under compression in the early phase of stance (86, 100). Functionally, Kleg is the summative lower-limb stiffness affecting performance during multi-jointed whole-body SSC actions.

3. *Joint stiffness* (Kjoint, $Nm \cdot rad^{-1}$) describes the angular variation of a joint in response to the (43).

4. *Musculotendinous stiffness* is calculated using the oscillation technique, in which an active and loaded muscle-tendon unit (MTU) is perturbed and its free resonance frequency response recorded (13).

Force and length are both factors of stiffness, thus, Kvert and Kleg are calculated during the period of ground contact, the former as the ratio of the peak vGRF to maximum COM displacement and the latter to peak leg compression (100). It is worth noting, however, that when the COM moves solely vertically, Kvert and Kleg are identical (14, 100). Yet, it has been proposed that Kvert is unable to account for the different contribution level of each joint in determining the whole leg's stiffness (62). Therefore, the concept of Kjoint was introduced, which models the relationship between joint moment and joint angle (156). Hence stiffness can be measured either for the whole-body system or for each joint in the system, but can also be measured passively, i.e. when muscles are not producing force.

As Kvert and Kleg exert influence over both spatiotemporal and kinematic variables, they are usually used in identifying these characteristics in individual runners. Prior research indicates that in SSC movements, Kjoint is the primary determinant of Kleg (4, 42, 44, 72, 76, 89), as joint angles on contact affected Kleg when runners performed such actions (42, 59, 72,

77, 101). Kleg is increased when a greater alignment of the vertical GRF vector relative to the joints reduces the moment arm of the GRF (115).

Whilst the relationship between lower-limb stiffness and athletic performance is seemingly logical, the evidence base is perhaps lacking and hence less definitive than it may be perceived by coaches and practitioners. Indeed, the literature reveals a considerable volume of inconsistencies. Previous review articles (13, 130) have attempted to identify the different measurements and methods by which to calculate lower-limb stiffness. Despite several methods for lower-limb stiffness analysis having been proposed (9, 24, 40, 41, 111, 113), Morin's sine-wave model (51) has been widely used for Kvert and Kleg determination due to its accuracy, efficacy, and the reduced amount of information it requires gathering (speed, leg length, contact time [CT], flight time [FT], and m).

Measurements of stiffness (see supplementary material)

Vertical stiffness

Vertical stiffness is generally accepted to be calculated as the quotient of maximum ground reaction force (GRF) and COM displacement (4, 33, 36, 39, 41, 46, 68, 69, 71, 111, 112, 115, 137). Kvert measured while performing activities which involve producing more force (e.g. running at higher velocities, single leg hopping as opposed to using both legs) was typically greater; however, measurement variation was also likely to be more diverse, which highlights the potential for reliability issues to arise when engaging in such tasks and the possible benefits of recruiting a larger sample. Data collected which indirectly found Kvert using the quotient of GRF and COM displacement formula produced results similar to studies where GRF and COM were directly measured, which suggests modelling those variables for measuring Kvert may provide a suitable alternative where direct measurement limitations exist. This is confirmed by Morin's study which recorded a small bias for results when GRF and COM displacement was modelled, as opposed to measured (111).

Leg stiffness

Leg stiffness is multifactorial in nature, reflecting influences of a range of active and passive musculoskeletal characteristics (43, 90). Previous studies indicate that Kleg is affected by hip, knee and ankle joint stiffness, to which both passive and active structures contribute (117). Then, Kleg measures lower-limb stiffness and is reliant upon leg compression which can only be achieved during stance (41, 69, 137). Although traditionally Kleg is assessed utilizing motion capture and inverse dynamics, this is constrained by the availability of sufficient training, time and cost. (14). Prior studies noted that Kleg was measured as the quotient of COM displacement and GRF during stance (27, 42, 44, 72, 75) (when, indeed, Kvert was calculated). However, the failure to consider flexion or extension at the hips or trunk, instead assuming a rigid body superior to the hips, constitutes a limitation in this study (130).

Where Kleg was investigated, it was most commonly measured using the quotient of GRF and change in leg length. Unfortunately, only three studies actually measured change in leg length (60, 125, 142), with most predicted it. As noted already, predicting rather than measuring leg length change is likely to negatively impact the accuracy of measurement. This is supported by Morin's study (111), which demonstrated that predicted change in leg length, although similar, is not exactly equivalent to measured change in leg length. Leg length changes at higher constant velocities (111, 112). It has been also shown that Kleg variation may increase due to greater CT (114).

Finally, Kleg is widely regarded as a substitute for loading rate and the subsequent kinematic response of the lower extremity during running. This applies where reduced Kleg is related to greater joint excursion and increased reliance on active muscle contributions to modulate landing tasks (107). Higher Kleg, meanwhile, is associated with reduced joint excursion and increased impulsive loading to bones and cartilage (151, 152). An association between heightened Kleg and incidents of lower extremity injury has been proposed (150); in

fact, Pruyn and colleagues (122) found that increased Kleg diversity between legs in Australian rules football players was also potentially linked to increased frequency of injury to lower extremities. Recurring association between the two could be indicative of a direct connection between Kleg and greater injury rates during dynamic activities.

Joint stiffness

Overall Kleg is governed by joint stiffness. Kjoint, defined as the proportional relationship between maximal joint moment and the maximum joint flexion in the middle of the stance phase (14), can be established via the torsional spring model (44). This model assumes that four rigid segments (foot, shank, thigh and head-arm-trunk) interconnect with torsional springs of the hip, knee and ankle and that lower extremity joints flex in the period between touch down and the middle of the ground contact phase.

Nevertheless, over the stance phase of running, muscle-tendon units (MTUs) around the ankle play a crucial role in the storage and generation of the energy necessary for propulsion (82, 129). The foot and ankle exhibit first a loading state, which entails the rising of the internal plantarflexor moment during dorsiflexion, and the absorption (and partial storage) of energy by the periarticular joint structures (52). This phase is followed by an unloading state in which the plantarflexion moment decreases while the joint plantarflexes, and the periarticular joint structures release their stored energy (52). This phase precedes a decline in the plantarflexion moment as the joint plantarflexes, at which time the periarticular joint structures release their stored energy, also known as an unloading state. The combination of MTUs surrounding the joint and the efferent neural motor pattern instantly controlling their mechanical characteristics contribute to ankle stiffness. (34, 45, 61). For instance, distinct features of muscle and tendon structural design in the lower limb have been shown to be present in habitual forefoot strikers; that, is, those who tend to land with a plantar-flexed ankle. Such architectural features include

(but are not limited to) shorter gastrocnemius medialis fascicles (25), thicker Achilles tendon (92) and stiffer foot arch (93).

Musculotendinous stiffness

To calculate MTU stiffness, the oscillation technique can be used (148, 149, 153, 154). Applying this technique, the free response to perturbing an active and loaded MTU was recorded as a means to model the human muscle as a damped spring system. Any disturbance to this system will result in damped oscillations (13). It is widely accepted that elastic energy storage and release in the SSC enhance the MTU's mechanical efficiency and power output (15) and it can therefore be concluded that SSC performance is susceptible to MTU elasticity (108). It has been suggested that this elasticity depends upon neuromuscular factors and intrinsic stiffness of MTU (10). Owing to the non-rigidity of the tendon, when the MTU lengthens, it is possible for electric energy to be retained. It is in this manner that SSC exercises can be enhanced by MTU performance (3, 15, 127). Regarding stiffness, there has been much emphasis on the MTU, and on tendon stiffness of knee extensors and plantar flexors in sprinting and running (72, 88, 142).

Passive stiffness

Thought it incorporates properties of other tissues, such as the skin, subcutaneous fat, fascia, ligament, joint capsule, and cartilage, passive stiffness has nonetheless been frequently used in evaluations of the mechanical properties of the MTU (84). Calculating passive stiffness is a relatively simple undertaking which has previously been determined via the use of an isokinetic dynamometer (48, 126, 144). For this calculation, once the subject is placed and secured in a dynamometer, the joint in question is taken through a range of motion. The subject avoided active resistance in order that passive resistance torque could be measured against angular displacement (48, 126) at hip, knee, and ankle joints.

In their earlier inquiry, Spurrs and colleagues (139) reported correlation between improved running economy by week 6 of plyometric training and enhanced passive stiffness in plantar flexors. Kubo et al. (87) posed on the basis of their findings that passive stiffness in the plantar flexors may in fact reflect muscle, rather than tendon, tissue. This was further supported by preceding research which concluded that passive plantar flexor stiffness corresponds to muscle stiffness measured using shear wave elastography (21, 70). Ueno and colleagues (144) stated that greater passive stiffness of the plantar flexors led to improved running performance in endurance runners, as their findings indicated significantly higher passive plantar flexor stiffness in well-trained endurance runners when compared with their untrained counterparts (144). Moreover, their results also showed that passive plantar flexor stiffness was higher in faster runners (144).

STIFFNESS AND RUNNING PERFORMANCE

Instigating a higher level of lower-limb stiffness is likely to be most beneficial in activities where the ability to transmit a given impulse more quickly would be advantageous, for instance, during maximum velocity running (11).

To describe stiffness in tasks such as hopping (73) and running (18), previous studies have employed the spring-mass model. To what extent the model can appropriately predict a task can be evaluated through calculation of the correlation coefficient between force and displacement. Conservative inclusion criteria ($r \geq 0.8$) has been applied to hopping investigations (57), an activity likely to be adequately described by the model as will be discussed below. However, it has been suggested that higher value ($r^2 \geq 0.9$) is more befitting when modelling sprinting gait and deviation from the spring-mass model (22).

Values for both Kvert and Kleg have been recorded during gait-based investigations, though the two may yield disparate results. Studies appeared to show that Kvert increases in line with running velocity (17, 50, 68, 89, 111, 112) and stride frequency (41). However, whilst

Arampatzis and colleagues found both Kvert and Kleg to increase with running velocity (4), still others demonstrated that Kleg does not in fact increase with running velocity (17, 68, 111). Such discrepancies could be taken to suggest that Kvert could be a more sensitive measure than Kleg when exploring relationships with running performance, a position which is also supported by further studies. For example, Morin and colleagues reported that fatigue-induced reductions in repeated sprint velocity were mirrored by reductions in Kvert, however, Kleg had no such influence on fatigue (112). Similar findings were reported regarding 800-m track running (55), and Nagahara and Zushi also found both greater Kvert values and increasing performance in sprinters after training, but no change in Kleg (116). Nevertheless, the reverse may be true where slower velocity, longer duration running is concerned; numerous studies reported reductions in Kleg and minimal change in Kvert following fatiguing protocols (30, 49, 67, 123, 124).

Morin and colleagues successfully demonstrated that, when ground CT was manipulated, fluctuations in the time spent in contact with the ground witnessed more variation in Kleg than in stride frequency (r^2 = 0.90 and 0.47, respectively) (114). Even though their research did not take into account the metabolic cost of running, an association between greater Kleg and diminished metabolic cost has been extracted from other studies (27) and therefore economical running strategy has been ascribed to this relationship (109). Consequently, it is arguable that producing a greater Kleg in conjunction with maintained stride frequency (facilitated by shorter CT) would result in a lowered metabolic cost of running. Additionally, a recent study identified that ground CT and Kleg are self-optimised characteristics while running (110). The authors reported that trained runners' performance met or approached their mathematical economical optimal in the process of submaximal running.

When comparing sprinting and endurance athletes while hopping (74) and completing 20-m progressive run and 30-m sprint (66), sprinters while hopping exhibited higher Kleg at

1.5 and 3.0 Hz and DJs from 30 cm, increased knee stiffness at 1.5 Hz, and greater ankle stiffness at 3.0. The aforementioned differences in Kleg and Kjoint between the two athletic groups has been attributed to heightened Achilles tendon stiffness observed in sprinters (5). However, it is also plausible that the greater Kleg reported for this group can be ascribed to more frequent strength and power training, increased relative strength capacity and greater SSC utilization, all of which are characteristic of sprinters.

In spite of showing lower Kleg than sprinters, endurance trained athletes exhibited higher Kleg than untrained subjects in hopping at 2.2 Hz (75). However, though the proposed explanation for greater Kleg in sprinters has been substantiated by existing literature, differences in Achilles and patellar tendon stiffness between endurance athletes and untrained subjects have been refuted (5, 65, 85, 128). Therefore, it can be postulated that increased Kleg in the endurance group may have been due to more prevalent slow-twitch muscle fibres, which is viewed to result from endurance training (78). For instance, when slow-twitch and fast-twitch muscle fibres were compared, the former demonstrated greater dynamic stiffness (121). Additionally, endurance training led to an increase in muscle stiffness, an increase associated with reduced fast-twitch muscle fibres (56).

The stiffer the leg, the more potentially effective its storage and release of energy, which may result in reducing the metabolic cost of running (110). Nonetheless, relationships have been captured between rises in both Kvert and Kleg, increased task intensity and improved task performance (98). During high velocity running tasks, Kvert may be more sensitive to change, whereas in exhaustive running Kleg may be more responsive to modifications (98).

STIFFNESS AND RUNNING-RELATED INJURIES

Running-related injuries are multifactorial. Runners can occur injury following repeated high-impact force without allowing sufficient time between application (79, 80). Implicated etiologic factors include: application of high force to the lower extremity tissues

during running (28, 79, 105); behavioural factors such as training history, injury history (23, 97, 99, 104, 143); and physiologic risk factors (35, 103). While there exists some evidence supports this general model of overuse injury (80), previous retrospective studies showed uninjured runners to have higher GRFs over injured runners (35, 103, 104, 106).

Messier and colleagues reported that the single significant means of injury prediction in their multivariate analysis was maximum knee stiffness, which was significantly higher in the injured group after training pace and body mass control protocols were put in place (106). In fact, both knee stiffness and body mass highly correlated within the injured group (106). Likewise, it has been suggested that higher knee stiffness, more common in runners with an increased body mass (≥80 kg), carries more possibility of suffering an overuse running injury (106). Given that knee stiffness involves aspects of force (knee extensor moment) and motion (knee flexion angle), it follows that a measure which incorporates both would be a feasible indicator of overuse running injuries.

INFLUENCING FACTORS ON LOWER-LIMB STIFFNESS

Many are the factors that may influence lower-limb stiffness while running (Figure 2). The way a runner's foot collides the ground, either the presence or absence of footwear, or the type of surface where one runs are just a few examples of stiffness influencing factors. Along this section, the most influencing elements affecting lower-limb stiffness during running are described as well as their relationship with Kvert and Kleg.

Foot-strike pattern

The FSP seems to correspond to behaviour of the various lower-limb stiffness assessments (62, 91, 94, 102, 131). A forefoot striking (FFS) pattern is defined as an FSP in which the ball of the foot connects with the ground ahead of the heel (94). It linked to the knee in respect of increased Kjoint and decreased range of motion (ROM), and to the ankle regarding lower Kjoint and higher ROM (156). These relationships are reversed for a rear-foot striking

(RFS) pattern (62, 91, 94, 102). The ratio of joint moment to joint angle ($\Delta M/\Delta\theta$) regulates these Kjoint measurements, which poses a potential mechanistic justification for the above observations; it could be that in terms of FFS, the increased ankle ROM leads to lowered ankle Kjoint, while the decreased knee ROM is responsible for the heightened knee Kjoint (156). Again, this is reversed where a RFS pattern is concerned (102). As previously stated, such interactions describe the effect of the FSP on Kjoint measurements. Since Kleg is more appropriate to describing the leg spring-like behaviour than the individual stiffness of each joint (156), it would be beneficial to analyse the affects of ankle and knee stiffness variations in relation to FSP type on the magnitude of global Kleg.

The literature appears to be divided on this matter (44, 91, 151). Farley and Morgenroth stated that Kleg is more sensitive to ankle Kjoint (44), selective sensitivity which is apparently attributable to the leg's geometry. Foot length oriented horizontally creates an extended GRF moment arm, which is associated with higher moment and angular displacements of the knee and hip joints (44). Thus, any RFS pattern which results definitively in ankle Kjoint increase would be more influential on the global Kleg (156). However, this experiment focused on a hopping task (44) which intrinsically involves a FFS pattern. On the other hand, others have held the knee to be the joint with greater influence on Kleg (91, 151). Contrary to Farley and Morgenroth's study where participants used a FFS pattern (44), in Williams and colleagues study all the runners exhibit a RFS pattern (151). This led them to use as their argument the simultaneous increase in knee Kjoint and Kleg in conjunction with the reduction in ankle Kjoint observed during the forefoot running experimental condition of the study (91). Apparently, Hamill and colleagues (62) were unique in testing ankle Kjoint deviance in two groups of runners with distinct FSP, classifying participants as either RFS or FFS runners. The groups were distinguished according to the presence of an impact peak on the vertical GRF and on the ankle angle at landing. Although runners may have been misclassified on the basis of these

criteria (53), nevertheless when running with their preferred FSP (FFS) exhibited a more compliant ankle and absorption of more negative work than habitual rearfoot strikers. Yet no differences were found when rearfoot strikers engaged their non-preferred mode, a FFS pattern (62).

FFS running seemingly augments the ability of the leg to store and reutilize elastic energy. However, the increased contractile costs resulting from increased muscle activation requirements, particularly in the triceps surge muscle group, counterbalance the mechanical advantages. Moreover, no differences can be observed between FFS and RFS patterns (120).

Footwear

A further influential factor in regulating lower-limb stiffness measures is footwear. Some runners exchanged their customary heel-cushioning shoes for minimalist shoes (141), a change with several biomechanical alteration implications. Firstly, as running barefoot tends to strike mid-foot or forefoot, it modifies stride length, which influences a range of factors: loading rate; plantar peak pressure; step frequency (SF) ; muscular activity; leg compliance; and ankle, knee and hip kinematics (29, 33, 94, 140, 155). Despite the tendency towards flatter foot placement on landing when transitioning from shod to barefoot running, there remain barefoot runners with heel-to-toe contact pattern (19).

Regarding the effects of footwear on lower extremity stiffness, studies have compared how Kleg differs when running without shoes against using traditional shoes (32). How Kleg (138) or Kjoint (6) is altered with different midsole hardness in traditional running footwear was also taken into account, in addition to how Kleg (96) or Kjoint (136) vary when running in minimalist versus traditional running shoes. Recent studies show that after a 4-week adaptation period, runners who wear fully minimalist shoes demonstrate higher Kvert and Kleg than runners using ultra-cushioning shoes (1). Cumulatively, these studies indicate clear impact of footwear on Kvert, Kleg, and Kjoint. Additionally, Jing and colleagues reported that while

shod Kvert, Kleg, and knee stiffness decrease, both hip and ankle stiffness increase in comparison with barefoot (83). Most of the existing literature centres on the effect of footwear condition on both Kleg and Kvert. In all the studies reviewed, the main finding was that barefoot running or running in minimalist footwear is accompanied by increases in Kleg (32, 33, 96, 131, 135). The recorded Kleg increases in both barefoot and minimally-shod running are attributable to either decreased leg compression through shorter CT, or increased vGRF values (156).

Another perspective is the interplay between Kvert and Kleg. Running barefoot or in minimal shoes was shown in two studies to precede an increase in Kleg but no significant alterations to Kvert (96, 131). This locomotor response is accredited to the body's adaptation strategy designed to prevent deviations from the customary displacement of the COM (46). Running with or without shoes, and the type of footwear where it is worn (i.e., minimalist vs. conventional), are stimuli that can alter the vertical displacement (46, 96). This will remain unchanged where Kleg adjustments are able to compensate for the imposed perturbations from the footwear condition (46, 96). It is worth noting that this observation (increased Kleg and unaltered Kvert) was not corroborated by Divert and colleagues, who reported instead simultaneous increases in Kvert and Kleg during the barefoot and minimalist running conditions (32, 33). It was put forward that the Kleg increase caused by barefoot running was not sufficient to maintain Kvert. The increase in Kvert was used to argue that barefoot running was superior in terms of energetic cost (32).

Ziliaskoudis and colleagues suggested that a greater vertical excursion in COM may occur in barefoot running compared to shod running and could therefore lead to increased total work production (156). The position that, due to shoe sole geometry (thicker at the heel, thinner under the footballs), the foot's heel is elevated with shoes, compared to a more horizontal foot orientation relative to the ground without, could be used to support Ziliaskoudis et al.'s findings

(94). Utz-Meagher and colleagues analysed precisely both barefoot and shod running (145). A FFS pattern, commonly adopted in barefoot running, also increases total work requirements as it increases joint excursions. Amplified plantarflexion at the ankle joint prior to initial ground contact forces the footballs to meet the ground first (145) before a dorsiflexion movement allows the heel to touch the ground, again followed by plantarflexion as the stance phase progresses (145). However, this initial lowering movement is partially absent in shod running due to the RFS pattern's prevalence (145). Shod running does not necessitate the higher work requirements imposed by the biomechanical characteristics of barefoot running, though it does not incur the metabolic penalty of the latter (156). The potential disadvantage for shod runners in terms of storage and return of elastic energy when compared to their barefoot peers is not translated into an oxygen consumption increase, due to lower total work demands (156).

It is arguable that technological shoe design evolution and advances in material development, particularly considering the established relationship between running economy and stiffness, could counterbalance the abovementioned diminished ability of storage and return of elastic energy at shod running compared to barefoot running (156).

Surface –type and slope

Runners modulate Kleg according to running surface. Kleg is lower on hard surfaces and higher Kleg on softer surfaces for their first step (46, 47). On softer surfaces, Kleg is increased to have the reverse effect on leg spring compression, which offsets the climb in surface compression to maintain the runner's COM path irrespective of surface stiffness. Because of the wide-reaching biomechanical parameters reliant upon the combined series stiffness of the runner and surface, Kleg modification allows humans to run in a similar manner on different surface stiffnesses (47). Stride frequency, ground CT, and peak GRF remain unaffected by surface stiffness (47). These observations are relevant to steady-state running on a continuous surface (46).

Little evidence is available on slope gradients and stiffness is available. García-Pinillos and colleagues studied the impact of numerous factors on spatiotemporal parameters during running: slope gradient, athletic level, Kvert and Kleg (51). Their findings on stiffness demonstrated that, relative to level running, Kvert increased with severe slopes (9-11%), whereas Kleg declined on moderate slopes (3-7%) regardless of the athletic level (51). Alternate research concluded that Kvert was raised during uphill running whilst Kleg remained constant across the different gradients included in the study (-8 to 8%) (96). The different methods in both studies may explain the slight differences between them. Whereas Lussiana required a pace of $10km \cdot h^{-1}$ over a range of slope gradients from -8 to 8%, García-Pinillos' study was executed at $12km \cdot h^{-1}$ using incremental slope gradients from 0-11%. Additionally, García-Pinillos and colleagues found notable correlation between Kleg and spatiotemporal parameters for level running, while Kvert was associated with spatiotemporal adaptations at more pronounced slope gradients (0-11%). The authors suggested that runners should engage greater force if they are to maintain velocity on steeper gradients resulting in increasing Kvert during running uphill (51).

Fatigue

How the spring-mass model behaves when a run is completed to exhaustion has not yet been clearly elaborated. García-Pinillos and colleagues (49), found that in practised runners, Kleg decreased while Kvert remained consistent, which is supported by previous work (36, 124). However, Hunter and Smith found Kleg or Kvert did not vary in participants who lacked training (81). Hayes and colleagues found that during a run to exhaustion both Kvert and Kleg decreased (67). Though negligible difference in Kvert was observed, the change in Kleg was significant and of a moderate magnitude over the course of the run to exhaustion (67). They also found that the maintenance of Kleg held strong associations both with the distance time period covered by the run to exhaustion. What is more, participant ability to maintain Kleg was

inversely proportionate to leg length change, CT and step length (SL) (67). A non-significant decline of modest magnitude in Kvert reflected the findings of prior work on fixed velocity runs (36, 124).

Regarding the relationship between velocity and stiffness, Enomoto and colleagues suggested that stiffness adjustment to running velocity is one of the key factors to keep pace in long-distance running (38). These authors stated that to acquire running velocity effectively, a runner should run with suitable Kvert. Furthermore, they proposed that if a runner has high Kvert even at low speed or keeps it despite the decrease in velocity, it might lead to fatigue and decrease in running velocity (38).

Velocity

Changes in lower-body stiffness over incremental-velocity protocol have already been reported (50, 111). The consequence of running at higher velocity are accepted to be increased step frequency, resulting in decreased CT, vertical displacement variation, as well as change in leg length (114). Fluctuation in both Kvert and Kleg produced by increasing velocity correlates with spatiotemporal running gait characteristics (50). Morin and colleagues determined that Kvert increases alongside increasing velocity while Kleg remains constant (111). García-Pinillos and colleagues found that high-level runners displayed greater stride angle and FT at high velocity (18 $km \cdot h^{-1}$) and SL (at 14-16-18 $km \cdot h^{-1}$) although, their amateur counterparts exhibited higher SF at 11-16-18 $km \cdot h^{-1}$ (50). Furthermore, amateur runners showed greater Kvert for all the velocities studied, whereas Kleg remained unchanged (50). It seems to be clear that Kvert increases as running velocity increases, while Kleg tends to remain unchanged.

Sex differences

Where controlled measurements of knee kinematics taken after mechanical perturbation during active flexion and extension exertions have recently been recorded and gender differences accounted for, women demonstrated less than 57% of the active muscle stiffness

compared to males (58). The Kleg is ascribed to the active muscle stiffness of the controlling joints (59) thereby affecting biomechanical stability. Granata and colleagues found that women exhibited lower Kleg than men in functional tasks while examining sex differences during two-legged hopping (57). Oscillation of the lighter female body mass necessitated the difference to facilitate equalling the hopping frequency of the heavier male subjects (57). However, during preferred hopping conditions, no constraints were put upon the female participants to employ lower stiffness (57). The women nevertheless consistently exhibited lower Kleg than the men, hopping at similar preferred frequencies and explanations for the mass-independent selection of this preference were proposed (57). Similarly, Padua and colleagues exposed reduced Kvert in women, but the gender difference was eliminated once the body mass was normalised (117), which explains Granata and colleagues' suggestion in the aforementioned study; sex differences in Kvert during a functional hopping task can be plausibly explained by anthropometric differences (57, 117). Nevertheless, Padua and colleagues found different stiffness recruitment between men and women revealing that female quadricep and soleus activity was significantly greater (117). Whilst the recruitment strategy may, in principle, efficiently modulate Kvert, it also has the potential to compromise knee joint stability. Particularly for women, oestrogen, aside from its familiar role as a sex hormone, is also a crucial factor in the development, maturation, and aging of extragonadal tissues such as bone (26, 64, 95), muscle (31, 37), and connective tissues (63, 64). There occurs natural variation in oestrogen secretion between young women, increasing 10- to 100-fold over the menstrual cycle (20). As the concentration of oestrogen rises during the menstrual cycle, so knee laxity rises, hence joint laxity has been found to be cyclical in nature (132-134). A change in knee laxity from 13.35 ± 2.53 mm during the follicular phase to 14.43 ± 2.60 mm during ovulation (118) resulted from a 17% reduction in knee stiffness during the ovulatory phase. Since the properties

of ligaments and tendons vary across the menstrual cycle, it should be considered while testing stiffness in women.

DISCUSSION AND IMPLICATIONS

Lower-limb stiffness is an outstanding concept when assessing and monitoring long-distance runners. Given its sensitivity, its assessment should be implemented considering slope gradient and running velocity. Practitioners are encouraged to develop non-fatiguing protocols where athletes execute the entire protocol wearing the same running shoes. Ultimately, both inter-athlete and between-sex comparisons are not recommended due to measurements sensitivity.

CONCLUSION

Given that lower-limb stiffness has been proved to be essential within spring-mass model's behaviour and its relation to both performance and injuries while running, this narrative review aimed to definitively identify influences on this feature. The topics discussed here have contributed to further solidify, increase, and broaden both the knowledge of lower-body stiffness behaviour and its scope of action. The effect of different variables on lower-limb stiffness should not be measured individually. It has been shown that one variable influences the others, thus, the behaviour of the spring-mass model should be analysed cautiously when altering factors such as foot-strike pattern or footwear. The findings reported here show that either the presence or absence of running shoes might contribute to the alteration of the FSP a runner adopts, what influences SF and, therefore, lower-limb stiffness at a given velocity. Despite both the SSC and lower-limb stiffness are key within the neuromuscular behaviour when elastic energy is used in sport, female athletes should be assessed cautiously as the menstrual cycle make musculotendinous properties fluctuate across it.

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10.4. Appendix 4. Ethics committee Study 1 and Study 3.

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GRUPO SANVALERO

Fecha 10/04/2018

INFORME Nº 009-17/18

COMITÉ ETICA UNIVERSIDAD SAN JORGE

Estudio	Autor
Influencia del contacto inicial, calzado y cadencia en los parámetros espacio-temporales de carrera.	**Luis Enrique Roche, Alejandro Almenar, José Antonio Bustillo, Gabriel Valiente, Felipe García-Pinillos, Pedro Latorre-Romón**

Características, Objetivos,

El objetivo de este estudio busca valorar las implicaciones y correlaciones existentes entre 3 variables que han demostrado influir en los patrones básicos de carrera (contacto inicial, condición de calzado y cadencia a velocidad conocida).
Metodológicamente se desarrollará un estudio transversal en 2 sesiones que explorará la interacción entre las 3 variables en el patrón de carrera realizado en tapiz rodante a velocidad confortable.

Resumen

El sujeto de estudio será sometido a los siguientes procedimientos:

Valoración antropométrica de los pies de los sujetos

- Baropodometría estática de los sujetos en plataforma de presiones.
- Medición de las características antropométricas del pie del sujeto en una plataforma de medición antropométrica (longitud, altura del arco al 50%, anchura del pie al 50%,...)

Medición de los parámetros espacio-temporales en tapiz rodante a velocidad confortable de 12 km/h variando 3 condiciones (calzado-descalzo, antepié-retropié, 150-160-170-180-190 pasos por minuto)

- Se usarán 5 sistemas (Optogait, Stryd, Runscribe, video-análisis, motion metrix) para analizar diferentes parámetros (longitud de zancada, ángulo de impulsión, tiempo de contacto y de vuelo, frecuencia de zancada...)
- Ninguna medición es invasiva y no suponen ningún riesgo para el sujeto.

Protocolo de medición

- Todas las mediciones se realizarán en dos sesiones previa citación con los investigadores: en una se medirán las características antropométricas y se realizará la prueba calzado, y en la otra se realizarán descalzos

- El paciente se equipará con su ropa habitual de carrera y las zapatillas habitualmente usadas para correr.

- El protocolo de medición se realizará de la siguiente manera:

  PRIMERA SESIÓN (duración estimada 1h/sujeto) :

  - La medición de las características antropométricas del pie se realizará al inicio en una tabla de mediciones con diferentes instrumentos calibrados. Se analizarán los siguientes datos: índice de altura del arco en carga y descarga, altura del escafoides relativa, largo del pie, anchura de antepié, mediopié y retropié y el índice de postura del pie.

  - Calentamiento de 10 minutos a velocidad confortable autoseleccionada.

  - Se realizará las mediciones (30 segundos cada una de ellas) con el sujeto calzado corriendo

    - Medición a 12 km/h en condición basal (autoseleccionada del paciente)
    - De antepié a 150-160-170-180-190 pasos/minuto
    - De retropié a 150-160-170-180-190 pasos/minuto
    - Medición a 12 km/h en condición basal (autoseleccionada del paciente)

  SEGUNDA SESIÓN (duración estimada 0.5 h/sujeto) :

  - Calentamiento de 10 minutos a velocidad confortable autoseleccionada.

  - Se realizará las mediciones (30 segundos cada una de ellas) con el sujeto descalzo corriendo

    - Medición a 12 km/h en condición basal calzado (autoseleccionada del paciente)
    - De antepié a 150-160-170-180-190 pasos/minuto
    - De retropié a 150-160-170-180-190 pasos/minuto
    - Medición a 12 km/h en condición basal calzado (autoseleccionada del paciente)

Lugar de realización del estudio

El estudio se realizará en el Laboratorio de Valoración Funcional ubicado en la planta baja del Edificio 3 de la Facultad de Salud de la Universidad San Jorge
Los días y fechas de exploración serán pautados concertados por los investigadores y

notificados con antelación suficiente.

Riesgos e inconvenientes
Los posibles riesgos e inconvenientes son casi nulos, ya que se basa en índices posturales, movimientos naturales y procedimientos habituales de carrera en tapiz rodante. Ninguna evaluación tiene carácter invasivo. No hay ningún tipo de riesgo para las personas en edad fértil. A su vez, no se modificará en ningún momento su pauta de entrenamiento ni carga habitual de actividad física.

Si el paciente presentara cualquier sintomatología durante el periodo de tiempo que dura el estudio, sería un criterio de abandono, con la posibilidad de volver a participar, si el paciente así lo deseara.

Medidas adoptadas

En base a la Ley 14/2007 de Investigación Biomédica, todos los participantes en dicho estudio deberán firmar un consentimiento informado para participar en este proyecto.

Toda la información personal obtenida y utilizada será tratada con confidencialidad siguiendo las directrices de la Ley Orgánica 15/1999 de Protección de Datos de Carácter Personal.

Resolución

A la vista de los datos aportados en relación al estudio, este Comité de Ética no observa disconformidad alguna para que se lleve a cabo en las condiciones que se nos indican.

En Villanueva de Gállego, a 19 de abril de 2018

Luis Carlos Correas
Presidente

10.5. Appendix 5. Informed consent Study 1 and Study 3.

CONSENTIMIENTO INFORMADO
DOCUMENTO DE INFORMACIÓN PARA EL PARTICPANTE
(De acuerdo con el Anexo XIV del Comité ético de Investigación Clínica de Aragón)

1.- CONSIDERACIONES GENERALES

Este documento sirve para que usted, o quien lo represente, dé su consentimiento para su participación en este proyecto de investigación. Eso significa que nos autoriza a realizar aquellos procedimientos necesarios para llevar a cabo el estudio.

Su participación es voluntaria y usted podrá revocar este consentimiento cuando lo desee. De su rechazo no se derivará ninguna consecuencia adversa respecto a la calidad del resto de la atención recibida. Antes de firmar, es importante que lea despacio la información siguiente.

Díganos si tiene alguna duda o necesita más información. Le atenderemos con mucho gusto.

2.- PROYECTO DE INVESTIGACIÓN

Influencia del contacto inicial, calzado, y cadencia en los parámetros espacio- temporales de la carrera.

3.- OBJETIVO Y MÉTODOS A UTILIZAR

El objetivo de este estudio es valorar las implicaciones y correlaciones existentes entre 3 variables que han demostrado influir en los patrones básicos de carrera (contacto inicial, condición de calzado y cadencia a velocidad conocida).

Metodológicamente se desarrollará un estudio transversal en 2 sesiones que explorará la interacción entre las 3 variables en el patrón de carrera realizado en tapiz rodante a velocidad confortable.

4.- PROCEDIMIENTOS A LOS QUE SE VA A SOMETER

Valoración antropométrica de los pies de los sujetos

- Baropodometría estática de los sujetos en plataforma de presiones.
- Medición de las características antropométricas del pie del sujeto en una plataforma de medición antropométrica (longitud, altura del arco al 50%, anchura del pie al 50%, ...)

Medición de los parámetros espacio-temporales en tapiz rodante a velocidad confortable de 12 km/h variando 3 condiciones (calzado-descalzo, antepié-retropié, 150-160-170-180-190 pasos por minuto)

- Se usarán 5 sistemas (Optogait, Stryd, Runscribe, video-análisis, motionmetrix) para analizar diferentes parámetros (longitud de zancada, ángulo de impulsión, tiempo de contacto y de vuelo, frecuencia de zancada…)
- Ninguna medición es invasiva y no suponen ningún riesgo para el sujeto.

Protocolo de medición

Todas las mediciones se realizarán en dos sesiones previa citación con los investigadores: en una se medirán las características antropométricas y se realizará la prueba calzado, y en la otra se realizarán descalzos

El paciente se equipará con su ropa habitual de carrera y las zapatillas habitualmente usadas para correr.

El protocolo de medición se realizará de la siguiente manera:

PRIMERA SESIÓN (duración estimada 1h/sujeto):

- La medición de las características antropométricas del pie se realizará al inicio en una tabla de mediciones con diferentes instrumentos calibrados. Se analizarán los siguientes datos: índice de altura del arco en carga y descarga, altura del escafoides relativa, largo del pie, anchura de antepié, mediopié y retropié y el índice de postura del pie.
- Calentamiento de 10 minutos a velocidad confortable autoseleccionada.
- Se realizará las mediciones (30 segundos cada una de ellas) con el sujeto calzado corriendo
    - Medición a 12 km/h en condición basal (autoseleccionada del paciente)
    - De antepié a 150-160-170-180-190 pasos/minuto
    - De retropié a 150-160-170-180-190 pasos/minuto
    - Medición a 12 km/h en condición basal (autoseleccionada del paciente).

SEGUNDA SESIÓN (duración estimada 0.5 h/sujeto):

- Calentamiento de 10 minutos a velocidad confortable autoseleccionada.
- Se realizará las mediciones (30 segundos cada una de ellas) con el sujeto descalzo corriendo
    - Medición a 12 km/h en condición basal calzado (autoseleccionada del paciente)
    - De antepié a 150-160-170-180-190 pasos/minuto
    - De retropié a 150-160-170-180-190 pasos/minuto

Riesgos e inconvenientes

Los posibles riesgos e inconvenientes son casi nulos, ya que se basa en índices posturales, saltos naturales y procedimientos habituales de carrera en tapiz rodante. Ninguna evaluación tiene carácter invasivo. No hay ningún tipo de riesgo para las personas en edad fértil. A su vez, no se modificará en ningún momento su pauta de entrenamiento ni carga habitual de actividad física.

Si el paciente presentara cualquier sintomatología durante el periodo de tiempo que dura el estudio, sería un criterio de abandono, con la posibilidad de volver a participar, si el paciente así lo deseara.

Lugar de realización del estudio

El estudio se realizará íntegramente en la universidad San Jorge

- Campus universitario Villanueva de Gállego, 50830, Villanueva de gallego, Zaragoza.

Los días y fechas de exploración serán concertados con los alumnos de fisioterapia Alejandro Almenar, José Antonio Bustillo, previo contacto por teléfono o correo electrónico.

Contraindicaciones.

Si tiene cualquier duda o consulta no dude en realizarla antes de comenzar:

- Sólo debe realizar estas pruebas físicas en el caso que su estado de salud sea acorde con la exigencia de las mismas. No debería realizarlas en el supuesto que:
    - Presente en la actualidad lesión traumatológica (ósea, muscular, tendinosa, ligamentosa).
- Presente en la actualidad lesión cardíaca o cardiocirculatoria de importancia, descompensada o no controlada (arritmia, problemas valvulares, insuficiencia cardíaca, hipertensión arterial).
- Se encuentre actualmente afecto por un proceso infeccioso agudo.
- No entienda el procedimiento a seguir para la realización de las pruebas físicas.

Otros procedimientos para los que pedimos su consentimiento

En algunos casos es necesaria la toma de imágenes, como fotos o videos. Sirven para documentar mejor el proceso. También pueden usarse para fines docentes o de difusión del conocimiento científico. En cualquier caso, las imágenes serán usadas sólo si usted da su autorización. Su identidad siempre será preservada de forma confidencial. En las imágenes las caras y señales de identidad serán difuminadas o tapadas de manera que impidan el reconocimiento.

5. AUTONOMÍA DEL PACIENTE

En todo momento, el paciente tendrá total libertad para revocar su participación (Ley 41/02 de Autonomía del Paciente), sin que su decisión influya negativamente en su posterior asistencia médica

La participación en este estudio tiene CARÁCTER VOLUNTARIO y de ninguna manera influirá en su atención médica

Puede llevarse la hoja de información a su casa para meditarla con tiempo suficiente y consultar su participación con su familia o con su médico habitual.

En caso de algún tipo de duda, aclaración o necesidad de una mayor información puede contactar con el **Investigador Principal Luis Enrique Roche** en el número 676637873 o bien con los fisioterapeutas

responsables de la intervención **Alejandro Almenar** (660228151), **José Antonio Bustillo** (653797266) y podólogo responsable de la intervención **Gabriel Valiente** (661111644).

6. FUENTE DE FINANCIACIÓN

El estudio actual cuenta con las siguientes fuentes de financiación:

- Universidad San Jorge
- Otras instituciones

7. USO DE LOS DATOS DERIVADOS DEL ESTUDIO

Si usted accede a colaborar en este estudio, debe saber que serán utilizados algunos datos sobre su salud, los cuales serán incorporados a una base de datos informatizada sin su nombre. Sus documentos médicos podrían ser revisados por personas dependientes de las
Autoridades Sanitarias, miembros de comités éticos independientes y otras personas designadas por ley para comprobar que el estudio se está llevando a cabo correctamente.

Sus datos serán objeto de un tratamiento disociado, vinculándose a un código, de modo que la información que se obtenga no pueda asociarse a persona identificada o identificable. Todos sus datos se mantendrán estrictamente confidenciales y exclusivamente el responsable del estudio conocerá su identidad. Los resultados del estudio podrán ser comunicados en reuniones científicas, congresos médicos o publicaciones científicas. En todo caso se mantendrá una estricta confidencialidad sobre la identidad de los pacientes. Se conservará en todo momento la confidencialidad personal sanitario-paciente (Ley de Protección de datos 15/1999).

Le informamos de que los datos de carácter personal recabados serán incorporados a un fichero titularidad de FUNDACION UNIVERSIDAD SAN JORGE, necesario para la correcta gestión del Proyecto de Investigación "Influencia del contacto inicial, calzado y cadencia en los parámetros espacio-temporales de carrera". Asimismo, le informamos de que solo se recogerán los datos estrictamente necesarios para la realización del mismo y que éstos no se comunicarán a terceros ajenos al Proyecto de Investigación, salvo en los supuestos legalmente previstos.

De acuerdo con lo dispuesto en la Ley Orgánica 15/1999, de Protección de Datos de Carácter Personal, en cualquier momento usted puede ejercitar sus derechos de acceso, rectificación, cancelación y oposición, enviando una solicitud por escrito acompaña de una fotocopia de documento oficial que lo identifique a la Universidad San Jorge, Autovía A-23 Zaragoza- Huesca, km. 299, 50830- Villanueva de Gállego (Zaragoza).

CONSENTIMIENTO INFORMADO

Ud. es libre de aceptar o no nuestra solicitud de participar en este proyecto. Si decide hacerlo, le rogamos que otorgue su consentimiento informado por escrito mediante la firma de este documento.

Título del proyecto de investigación
“Influencia del contacto inicial, calzado y cadencia en los parámetros espacio-temporales de carrera.”

Yo, _ con NIP_ _ _ _ _ _ _ _ _ _ _

☐ He leído la hoja de información que se me ha entregado

☐ He sido informado de forma clara, precisa y suficiente de los extremos que afectan a los datos personales que se contienen en este consentimiento y en la ficha o expediente que se abra para la realización del Proyecto de investigación.

☐ He podido hacer preguntas sobre el estudio y he recibido suficiente información sobre el mismo.

☐He hablado con _

Comprendo que mi participación es voluntaria.

Comprendo que puedo retirarme del estudio:
1) cuando quiera
2) sin tener que dar explicaciones
3) sin que esto repercuta sobre mi persona

A continuación se detallan los supuestos en los que usted puede manifestar su negativa al tratamiento, uso y publicación de sus datos personales, muestras biológicas y pruebas físicas recabados para la realización del Proyecto citado, según ha sido debidamente informado, los cuales están incorporados a un fichero titularidad de la FUNDACION UNIVERSIDAD SAN JORGE con la única finalidad del correcto desarrollo del presente Proyecto de Investigación: Influencia del contacto inicial, calzado y cadencia en los parámetros espacio-temporales de carrera.”

☐ Si lleva a cabo la marcación de esta casilla, usted presta consentimiento al tratamiento de sus datos personales y pruebas físicas con fines estadísticos y científicos, lo cual se llevará a cabo mediante procesos adecuados de disociación de datos que impidan su identificación.

☐ Si lleva a cabo la marcación de esta casilla, usted presta consentimiento al tratamiento de sus datos personales y pruebas físicos con fines de investigación, lo cual se llevará a cabo siempre mediante procesos adecuados de disociación de los datos que impidan su identificación.

☐ Si lleva a cabo la marcación de esta casilla, usted presta consentimiento a la publicación de los resultados de investigación, resultados estadísticos o científicos, publicación que únicamente reflejará datos disociados que no permitan la identificación de los participantes en el Proyecto de Investigación.

☐ Si lleva a cabo la marcación de esta casilla, usted presta consentimiento al tratamiento de sus datos personales y exámenes físicos con fines docentes, lo cual se llevará a cabo siempre mediante procesos adecuados de disociación de los datos que impidan su identificación.

☐ Si lleva a cabo la marcación de esta casilla, usted presta consentimiento a la toma de imágenes (fotos y/o vídeos) a efectos de documentar el caso durante la realización del Estudio.

☐ Si lleva a cabo la marcación de esta casilla, usted presta consentimiento al uso de las imágenes tomadas (fotos y/o vídeos) durante la realización del Estudio, para fines docentes de difusión del conocimiento científico del presente Proyecto de Investigación.

☐ Si lleva a cabo la marcación de esta casilla, usted presta a que sus datos clínicos sean revisados por personal ajeno al centro con la única finalidad de la realización del presente Proyecto, de conformidad con la normativa vigente en materia de Protección de Datos.

☐ Si lleva a cabo la marcación de esta casilla, usted presta consentimiento a que las muestras derivadas de este estudio sean utilizadas en futuras investigaciones relacionadas con ésta.

Con la firma del presente documento, y si realiza la marcación de las casillas correspondientes, usted otorga consentimiento al tratamiento de los datos personales, exámenes físicos e imágenes que nos ha proporcionado como participante en el Proyecto “Influencia del contacto inicial, calzado y cadencia en los parámetros espacio-temporales de carrera.”, que podrá ser revocado en cualquier momento sin que de ello se derive consecuencia alguna para usted.

Deseo ser informado sobre los resultados del estudio: SÍ NO

He recibido una copia firmada de este Consentimiento Informado.

Firma del participante y fecha Firma del investigador y fecha

DENEGACIÓN O REVOCACIÓN DE CONSENTIMIENTO

Después de ser informado de la naturaleza y riesgos del procedimiento propuesto, manifiesto de forma libre y consciente mí:

DENEGACIÓN/REVOCACIÓN DE CONSENTIMIENTO para su realización, haciéndome responsable de las consecuencias que pueden derivarse de esta decisión.

Firma del participante y fecha Firma del investigador y fecha

10.6. Appendix 6. Ethics committee Study 2.

universidad SANJORGE GRUPO SANVALERO

Fecha 26/02/2019

INFORME Nº 006-18/19

COMITÉ ETICA UNIVERSIDAD SAN JORGE

Estudio	Autor
RIGIDEZ DE LA EXTREMIDAD INFERIOR EN CORREDORES DE RESISTENCIA	**Luis Enrique Roche Seruendo** **Felipe García Pinillos (Univ Jaén)** **Antonio Cartón Llorente** **Diego Jaén Carrillo** **Alejandro Almenar** **Amandine Gonthier** **Alberto Rubio Peirotén**

Características, Objetivos,

Objetivo primario: Establecer la correlación entre la rigidez del tren inferior en corredores de resistencia con los parámetros espaciotemporales.

Objetivos secundarios:

- Comparar las características del tendón de Aquiles, fascia plantar y tendón rotuliano en diferentes poblaciones de corredores de diferentes niveles de rendimiento.
- Valorar la relación entre el grosor de los tendones implicados en salto y carrera con los parámetros espaciotemporales.

Resumen

MATERIAL Y MÉTODO

PROTOCOLO PRE-CARRERA

- Antropometría
  - Se medirá la altura y peso de los sujetos, así como composición corporal mediante TANITA.
  - De manera específica se medirán las características antropométricas del pie del sujeto mediante dos sistemas de medida
  - Plataforma antropométrica del pie: Es una plataforma sobre la cual se realizan medidas anatómicas del pie cómo la longitud, la altura del arco en diferentes condiciones, la anchura de la zona media del pie, etc.
    - Estas mediciones se harán con el protocolo Sit-to-Stand. Se repetirán las mismas mediciones con el paciente en sedestación, en bipedestación sobre el pie derecho para valorar el grado de estabilidad o flexibilidad del pie al someterse a la carga.
  - Plataforma baropodométrica: Medición de las presiones plantares a través de una plataforma sensorizada. Se realizará mediciones en bipedestación estática durante 10 segundos y dinámica caminando sobre la misma.

- Personal entrenado realizará la valoración del índice de postura del pie de 6 ítems es un índice clínico utilizado de manera habitual en la podología y traumatología para la caracterización del pie. Este índice ha mostrado correlación con la estructura anatómica y la función del pie. La puntuación final del índice se consigue mediante la valoración, manual o visual, de 6 ítems por parte de un profesional entrenado.

- Cuestionario sociodemográfico y de rendimiento.
- Caracterización mediante ecografía con el dispositivo LOGIQ S7 EXPERT (General Electric, Alemania, 2013) y el transductor lineal ML 6-15MHz. MATRIX LINEAR. Todos los parámetros de configuración del sistema se mantendrán en Modo B y constante para cada una de las siguientes estructuras analizadas:
  - Tendón Aquiles:
    - Se utilizará una frecuencia de 12Mhz, una profundidad de 2 cm, el foco a 0,5 cm y una ganancia de 100dB.
      Para la realización de la valoración ecográfica el sujeto estará en decúbito prono, con los tobillos en posición neutra y los pies fuera de la camilla. El grosor del tendón se medirá mediante una toma longitudinal en la referencia situada 3 cm proximal a la inserción del tendón en el calcáneo. Este punto se identificará mediante una marca en la piel (Del Bano-Aledo et. al, 2017).
  - Tendón rotuliano:
    - Se utilizará una frecuencia de 12Mhz, una profundidad de 3 cm, el foco a 0,5 cm y una ganancia de 100dB.
      Para la realización de la valoración ecográfica el sujeto estará en decúbito supino, con ambas rodillas flexionadas a 30°. El grosor del tendón se medirá mediante una toma longitudinal en la referencia situada 1 cm distal al polo inferior de la rótula. Este punto se identificará mediante una marca en la piel (Del Bano-Aledo et. al, 2017).
  - Fascia plantar:
    - Se utilizará una frecuencia de 12Mhz, una profundidad de 3 cm, el foco a 1 cm y una ganancia de 100dB.
      Para la realización de la valoración ecográfica el sujeto estará en decúbito prono, con los tobillos en posición neutra y los dedos extendidos contra la superficie de la camilla de valoración. El grosor de la fascia plantar se medirá mediante una toma longitudinal en la referencia situada desde el borde anterior de la superficie plantar del calcáneo verticalmente hasta el borde inferior de la fascia plantar. Este punto se identificará mediante una marca en la piel (Del Bano-Aledo et. al, 2017).
- Calentamiento en tapiz rodante durante 7' con incremento de velocidad de 1km/h hasta alcanzar una velocidad cómoda para el sujeto la cual se mantendrá durante 3'. Posteriormente se realizará otra toma de 3 minutos con el sujeto descalzo con 2 minutos para el cambio y adaptación.
- Realización de un drop jump desde una altura de 20cm y 30cm, el cuál será grabado mediante el sistema Optogait. Cada sujeto realizará tres intentos, teniendo validez para el estudio la mejor marca obtenida.

PROTOCOLO DE CARRERA

Se realizará una prueba de carrera de 7'+3'+3' en tapiz rodante HP Cosmos. La velocidad inicial será dispuesta a 8 Km/h y habrá un incremento gradual de velocidad hasta alcanzar una velocidad confortable para el sujeto. Durante la prueba, se controlarán las variables de calzado, tipo de contacto y cadencia del corredor.

MEDICIONES DURANTE EL PROTOCOLO DE CARRERA:

- STRYD: potenciómetro de carrera
- Runscribe cada bloque: percepción subjetiva al esfuerzo
- Optogait: medición de parámetros espaciotemporales de carrera
- MotionMetrix: captura del movimiento en 3D sin marcadores para análisis biomecánico de la carrera

PROTOCOLO POST-CARRERA:

- 2' de decremento gradual de carrera hasta finalizar la prueba
- Fin de la prueba

CRITERIOS DE INCLUSIÓN:

Corredores recreacionales que quieran participar libremente en el estudio y que cumplan todos los criterios que se mencionan a continuación:

- Haber corrido sobre un tapiz rodante al menos 5 veces una distancia superior a 1 km.
- Experiencia previa de entrenamiento 1 año
- Mejor marca en carrera de 10 km. con un tiempo comprendido entre los 30 y los 60 minutos
- Experiencia previa de entrenamiento de un año, realizando durante los últimos 6 meses, al menos, dos entrenamientos de carrera a la semana
- Hombres y mujeres sanos: que no presenten ninguna enfermedad cardiovascular ni lesiones durante los últimos 3 meses pudiendo así garantizar el entrenamiento.
- Deben presentar una prueba de esfuerzo realizada durante el último año que acredite que no hay ningún problema cardiovascular.

CRITERIOS DE EXCLUSIÓN

- Presentar alguna patología neurológica, musculo-esquelética, cardiaca o sistémica que impida la práctica deportiva

CRITERIOS DE ABANDONO

- Se trata de un estudio observacional con una duración estimada de 30 minutos donde las pruebas a las que se somete al sujeto no exigen ninguna actividad de riesgo a las que no esté habituado.
- Si el sujeto presentara cualquier sintomatología durante la realización de las pruebas del estudio, se consideraría criterio de abandono con la posibilidad de volver a participar en otra sesión si así lo deseara.

Todos los sujetos serán informados previamente de los objetivos del estudio y de las intervenciones a las que serán sometidos mediante un consentimiento informado, el cual leerán y firmarán voluntariamente.

En la anamnesis del paciente se recogerán los datos que aparecen en el anexo "Cuestionario sociodemográfico"

Medidas adoptadas

El desarrollo del proyecto se basará en las Declaraciones de la Asociación Médica Mundial del Helsinki. Se informará a cada sujeto acerca de la naturaleza del estudio, voluntariedad de la participación en el mismo, de los objetivos propuestos, así como de los posibles efectos adversos que pudieran tener lugar en su realización. A cada sujeto se le solicitará que dé su consentimiento a participar en el estudio por escrito. El estudio será suspendido en cualquier momento, si así lo desea el paciente.

De acuerdo con lo establecido en el artículo 7 de la Ley 41/2002, así como en el artículo 7.3 de la Ley orgánica 15/1999, de 13 de diciembre de protección de datos de carácter personal y en el Reglamento (UE) 2016/679 del Parlamento Europeo y del Consejo de 27 de abril de 2016, que comenzó a aplicarse el 25 de mayo de 2018, se respetará rigurosamente la confidencialidad de los datos de carácter personal y de salud de los participantes.

Los datos de los sujetos de estudios serán codificados en un archivo diseñado para tal fin por un investigador que no participe en las mediciones, sin incluirse ningún dato que permita la identificación de los mismos.

Resolución Comité Ética

A la vista de los datos y documentación adicional aportados en relación al estudio, este Comité de Ética no observa disconformidad alguna para que se lleve a cabo en las condiciones que se nos indican.

En Villanueva de Gállego, a 26 de febrero de 2019

29103861Y LUIS CARLOS CORREAS (R: G99047672)

Firmado digitalmente por 29103861Y LUIS CARLOS CORREAS (R: G99047672)
Fecha: 2019.02.26 22:33:29 +01'00'

Luis Carlos Correas
Presidente

10.7. Appendix 7. Informed consent Study 2.

CONSENTIMIENTO INFORMADO
DOCUMENTO DE INFORMACIÓN PARA EL PARTICPANTE
(De acuerdo con el Anexo XIV del Comité ético de Investigación Clínica de Aragón)

1.- CONSIDERACIONES GENERALES

Este documento sirve para que usted, o quien lo represente, dé su consentimiento para su participación en este proyecto de investigación. Eso significa que nos autoriza a realizar aquellos procedimientos necesarios para llevar a cabo el estudio.
Su participación es voluntaria y usted podrá revocar este consentimiento cuando lo desee. De su rechazo no se derivará ninguna consecuencia adversa respecto a la calidad del resto de la atención recibida. Antes de firmar, es importante que lea despacio la información siguiente.

Díganos si tiene alguna duda o necesita más información. Le atenderemos con mucho gusto.

2.- PROYECTO DE INVESTIGACIÓN
"Rigidez de la extremidad inferior en corredores de resistencia".

3.- OBJETIVO Y MÉTODOS A UTILIZAR

El objetivo de este estudio es establecer la correlación entre la rigidez del tren inferior en corredores de resistencia con los parámetros espaciotemporales de carrera comparando las características del tendón de Aquiles, fascia plantar y tendón rotuliano en diferentes poblaciones de corredores de diferentes de niveles de rendimiento.

Estos datos servirán de soporte a futuras investigaciones y planes de prevención de tendinopatías, poner de manifiesto la función y características de las diferentes estructuras y tejidos del tren inferior y su influencia en procesos patológicos y rendimiento deportivo.

4.- PROCEDIMIENTOS A LOS QUE SE VA A SOMETER

Valoración antropométrica del sujeto
Se medirá la altura y peso de los sujetos, así como composición corporal mediante TANITA.
De manera específica se medirán las características antropométricas del pie del sujeto medidas con un sistema de plataforma antropométrica del pie

- Es una plataforma sobre la cual se realizan medidas anatómicas del pie cómo la longitud, la altura del arco en diferentes condiciones, la anchura de la zona media del pie, etc. a través de pies de rey electrónicos
- Medición estática y dinámica de las presiones plantares con plataforma baropodométrica

Valoración postural del pie del sujeto mediante el "Índice de postura del pie" (FPI-6)

El índice de postura del pie de 6 ítems es un índice clínico utilizado de manera habitual en la podología y traumatología para la caracterización del pie. Este índice ha mostrado correlación con la estructura anatómica y la función del pie.
La puntuación final del índice se consigue mediante la valoración, manual o visual, de 6 ítems por parte de un profesional entrenado.

Caracterización mediante ecografía del tendón de Aquiles, tendón rotuliano y fascia plantar

- El paciente se pondrá en pantalón corto en decúbito en la camilla.
- Se procederá a la aplicación de gel de ultrasonidos y control ecográfico.
- Se evaluará exclusivamente el lado derecho.
- La evaluación se realizará previamente a la exploración en carrera para no implicar carga o modificación alguna en los tendones respecto a la situación de reposo.

Valoración de los parámetros espaciotemporales de la marcha con sistemas optoeléctricos (Optogait) y sistemas markerless (sin marcadores) en tapiz rodante

- El paciente se equipará con su ropa habitual de carrera y las zapatillas habitualmente usadas para correr. Se pedirá el uso de calcetín grueso y vaselina para la parte de la prueba en la que se corre descalzo.
- Se realizará un calentamiento general de duración aproximada de 7 minutos sobre un tapiz rodante.
- Tras el calentamiento se comenzará la prueba a una velocidad establecida por el paciente superior a 10 km/h pero que sea confortable para proceder a la medición.
- El protocolo de medición se realizará de la siguiente manera:

  Se realizará una prueba de carrera de 10' en tapiz rodante HP Cosmos. La velocidad inicial será dispuesta a 8 Km/h y habrá un incremento gradual de velocidad hasta alcanzar la velocidad confortable de carrera que será mantenida durante 3 minutos. Posteriormente se pedirá al paciente que corra durante 3 minutos descalzo (simplemente usando unos calcetines y protegiendo la piel con una crema a base de vaselina). Durante la prueba, se controlarán las variables de calzado, tipo de contacto, velocidad y cadencia del corredor.

Valoración de la capacidad de salto con sistemas optoeléctricos.

- Posteriormente se realizarán tres Drop Jump (DJ) se trata de un salto con caída y rebote desde una altura de 20 y 30 cm con un sistema Optogait:
  - 3x Drop Jump 20 cm: salto desde una altura de 20 cm con rebote pliométrico.
  - 3x Drop Jump 30 cm: salto desde una altura de 30 cm con rebote pliométrico.

Riesgos e inconvenientes

Los posibles riesgos e inconvenientes son menores y se relatan a continuación.

- Ninguna intervención o evaluación tiene carácter invasivo.
- No hay ningún tipo de riesgo para las personas en edad fértil.
- A su vez, no se modificará en ningún momento su pauta de entrenamiento ni carga habitual de actividad física.
- Los riesgos son:
  - Aparición de flictenas o ampollas en la dermis de la planta del pie. Se recomienda el uso de calcetín grueso y crema con base de vaselina (que se provee por parte del equipo investigador) para evitar su aparición.

Si el paciente presentara cualquier sintomatología durante el periodo de tiempo que dura el estudio, sería un criterio de abandono, con la posibilidad de volver a participar, si el paciente así lo deseara.

Lugar de realización del estudio

El estudio se realizará en el Laboratorio de Valoración Funcional ubicado en la planta baja del Edificio 3 de la Facultad de Salud de la Universidad San Jorge

Los días y fechas de exploración serán pautados y concertados por los investigadores y notificados con antelación suficiente.

Contraindicaciones.

Si tiene cualquier duda o consulta no dude en realizarla antes de comenzar:

- Sólo debe realizar estas pruebas físicas en el caso que su estado de salud sea acorde con la exigencia de las mismas. No debería realizarlas en el supuesto que:

- Presente en la actualidad o haya sufrido en los últimos 6 meses una lesión traumatológica (ósea, muscular, tendinosa, ligamentosa).
- Presente en la actualidad lesión cardíaca o cardiocirculatoria de importancia, descompensada o no controlada (arritmia, problemas valvulares, insuficiencia cardíaca, hipertensión arterial).
- Presente en la actualidad lesión que curse con mareos, vértigos, o inestabilidad de cualquier tipo.
- Se encuentre actualmente afecto por un proceso infeccioso agudo.
- No entienda el procedimiento a seguir para la realización de las pruebas físicas.

Otros procedimientos para los que pedimos su consentimiento
En algunos casos es necesaria la toma de imágenes, como fotos o videos. Sirven para documentar mejor el proceso. También pueden usarse para fines docentes o de difusión del conocimiento científico. En cualquier caso, las imágenes serán usadas sólo si usted da su autorización. Su identidad siempre será preservada de forma confidencial. En las imágenes las caras y señales de identidad serán difuminadas o tapadas de manera que impidan el reconocimiento.

5. FUENTE DE FINANCIACIÓN
El estudio actual cuenta con las siguientes fuentes de financiación:

- Universidad San Jorge

6. AUTONOMÍA DEL PACIENTE

En todo momento, el paciente tendrá total libertad para revocar su participación (Ley 41/02 de Autonomía del Paciente), sin que su decisión influya negativamente en su posterior asistencia médica La participación en este estudio tiene **CARÁCTER VOLUNTARIO** y de ninguna manera influirá en su atención médica.

Puede llevarse la hoja de información a su casa para meditarla con tiempo suficiente y consultar su participación con su familia o con su médico habitual.

Al tratarse de un estudio con participación de alumnos de la propia Universidad San Jorge debemos reincidir y destacar que:

- La participación tiene un **CARÁCTER VOLUNTARIO**
- No existirá ningún beneficio ni perjuicio académico directo o indirecto derivado de la participación o no participación como sujeto de estudio.
- La participación no afectará en ningún momento el normal funcionamiento de las clases ni intercederá en bajo ningún concepto en el proceso académico del sujeto de estudio.
- Todos los datos serán estrictamente encriptados y disociados garantizando la confidencialidad personal sanitario-paciente Reglamento (UE) 2016/679 del Parlamento Europeo y del Consejo de 27 de abril de 2016 de Protección de Datos (RGPD).
- En este estudio **NO** se tomarán datos considerados de alto nivel de protección o susceptibles de generar discriminación o estigmatización social o perjuicio personal o familiar (Ideología política o religiosa, vida sexual, actividades ilegales o antisociales, consumo de alcohol o drogas, enfermedades mentales o problemas psicológicos graves, datos sobre conductas de discriminación o acoso activo o pasivo, maltrato o abuso físico, psíquico o sexual activo o pasivo)

En caso de algún tipo de duda, aclaración o necesidad de una mayor información puede contactar con el Investigador Principal **Luis Enrique Roche** en el número **676637873** o en el email **leroche@usj.es**.

7.- USO DE LOS DATOS DERIVADOS DEL ESTUDIO

Si usted accede a colaborar en este estudio, debe saber que serán utilizados algunos datos sobre su salud, los cuales serán incorporados a una base de datos informatizada sin su nombre. Sus documentos médicos podrían ser revisados por personas dependientes de las
Autoridades Sanitarias, miembros de comités éticos independientes y otras personas designadas por ley para comprobar que el estudio se está llevando a cabo correctamente.
Sus datos serán objeto de un tratamiento disociado, vinculándose a un código, de modo que la información que se obtenga no pueda asociarse a persona identificada o identificable. Todos sus datos se mantendrán estrictamente confidenciales y exclusivamente el responsable del estudio conocerá su identidad. Los resultados del estudio podrán ser comunicados en reuniones científicas, congresos médicos o publicaciones científicas. En todo caso se mantendrá una estricta confidencialidad sobre la identidad de los pacientes. Se conservará en todo momento la confidencialidad personal sanitario-paciente (Reglamento (UE) 2016/679 del Parlamento Europeo y del Consejo de 27 de abril de 2016 de Protección de Datos (RGPD).
De acuerdo con dicha normativa, le informamos de que el responsable del tratamiento de los datos personales será FUNDACION UNIVERSIDAD SAN JORGE. Asimismo, le informamos de que solo se recogerán los datos estrictamente necesarios para la realización del mismo y que éstos no se comunicarán a terceros ajenos al Proyecto de Investigación, salvo en los supuestos legalmente previstos.
Como participante en el estudio puede ejercitar sus derechos de acceso, modificación, oposición, cancelación, limitación del tratamiento y portabilidad, dirigiéndose al Delegado de Protección de Datos de la Universidad adjuntando a su solicitud de ejercicio de derechos una fotocopia de su DNI o equivalente al domicilio social de USJ sito en Autovía A-23 Zaragoza- Huesca, km. 299, 50830-Villanueva de Gállego (Zaragoza), o la dirección de correo electrónico privacidad@usj.es. Asimismo, tiene derecho a dirigirse a la Agencia Española de Protección de Datos en caso de no ver correctamente atendido el ejercicio de sus derechos.

8. CONSENTIMIENTO INFORMADO

Ud. es libre de aceptar o no nuestra solicitud de participar en este proyecto. Si decide hacerlo, le rogamos que otorgue su consentimiento informado por escrito mediante la firma de este documento

Título del proyecto de investigación
“Rigidez de la extremidad inferior en corredores de resistencia”

Yo, _, con NIP_ _ _ _ _ _ _ _ _ _,

☐ He leído la hoja de información que se me ha entregado

☐ He sido informado de forma clara, precisa y suficiente de los extremos que afectan a los datos personales que se contienen en este consentimiento y en la ficha o expediente que se abra para la realización del Proyecto de investigación.

☐ He podido hacer preguntas sobre el estudio y he recibido suficiente información sobre el mismo.

☐He hablado con _

Comprendo que mi participación es voluntaria: SÍ NO

Comprendo que puedo retirarme del estudio:
1) cuando quiera
2) sin tener que dar explicaciones
3) sin que esto repercuta sobre mi persona

Como participante podrá retirarse del estudio en cualquier momento comunicándoselo al investigador principal, si bien queda informado de que sus datos no podrán ser eliminados para garantizar la validez de la investigación y garantizar el cumplimiento de los deberes legales del responsable.
Igualmente queda informado de que los resultados del presente proyecto podrán ser usados en el futuro en otros proyectos de investigación relacionados con el campo de estudio objeto del presente, así como que tiene derecho a ser informado sobre los resultados del estudio en el caso de que así lo solicite.

Deseo ser informado sobre los resultados del estudio: SÍ NO

He recibido una copia firmada de este Consentimiento Informado.

Firma del participante y fecha Firma del investigador y fecha

Firma de los padres o tutores (si procede)* y fecha
*Procede el consentimiento paterno en personas menores de 14 años

DENEGACIÓN O REVOCACIÓN DE CONSENTIMIENTO

Después de ser informado de la naturaleza y riesgos del procedimiento propuesto, manifiesto de forma libre y consciente mí:

- DENEGACIÓN/REVOCACIÓN DE CONSENTIMIENTO para su realización, haciéndome responsable de las consecuencias que pueden derivarse de esta decisión.

Firma del participante y fecha

Firma del investigador y fecha

Firma de los padres o tutores (si procede)* y fecha

UNIVERSIDAD SAN JORGE
Facultad de Ciencias de la Salud

"The human body contains all traits that are perfect for endurance and ultra-endurance sports: no body fur and a glut of sweat glands that keep us cool while running; narrow waists and long legs compared to our frames; large surface areas of joints for shock absorption. We have an arch in our foot that acts like a spring, short toes that are better for pushing off than for grasping tree limbs. When we run we can turn our torso and our shoulders while keeping our head straight and we have big butt muscles that keep us upright while running"

David Epstein