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## A State-of-the-art Review on the Underlying Mechanisms of Running Economy

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**A State-of-the-art Review on the Underlying Mechanisms of Running Economy**

By

Paul Allen Sage

Accepted in Partial Completion  
of the Requirements for the Degree  
Master of Science

ADVISORY COMMITTEE

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## Master's Thesis

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Paul A. Sage

July 11, 2022

**A State-of-the-art Review on the Underlying Mechanisms of Running Economy**

A Thesis

Presented to

The Faculty of

Western Washington University

In Partial Fulfilment

Of the Requirement for the Degree

Master of Science

By

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## **Abstract**

Running economy is determined by multiple physiological, biomechanical, and neuromuscular variables. Lower extremity stiffness has been identified as a primary factor in the determination of running economy due to its role in the utilization of elastic energy. Recent research exploring fascia has uncovered new insights into the underlying mechanisms driving the storage and utilization of elastic energy during running. New insights coupled with improvements in musculoskeletal modeling techniques and in instrumentation, allowing for accurate non-invasive quantification of passive stiffness, has opened the door for further exploration of running economy. The intention of this state-of-the-art review is to summarize the established conventions regarding the underlying mechanisms of running economy, and specifically stiffness, while reviewing current advancements in research on the topic to identify potential directions for future primary research. An electronic search of PubMed, Google Scholar, Semantic Scholar, and Mendeley databases was conducted, and the results were screened. Forward and backward citations were evaluated, and relevant literature was included to establish context, scope, and an accurate chronology of the development of the topic. A review of the relevant literature highlighted the crucial role that the modulation of stiffness plays in running economy and revealed strong evidence that fascia plays a significant role in the utilization of elastic energy. Additionally, the shortcomings of previous research and methodology was evaluated revealing avenues for further exploration of running economy.

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This work is dedicated to the memory of an incredible father, friend, and physicist Dr. Keith A. Sage.

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## 1. Introduction

Running performance is determined by the combination of maximal oxygen consumption ( $\text{VO}_2\text{max}$ ), fractional utilization of  $\text{VO}_2\text{max}$  ( $\% \text{VO}_2\text{max}$ ), anaerobic threshold (LT) and running economy (RE) (82,83,139). Running economy, defined as the steady-state oxygen consumption ( $\text{VO}_2$ ) at a given submaximal speed, is determined by the interaction of multiple physiological and biomechanical factors (46,128,131,132). Running economy functions as a key determining factor for distance running performance in that it dictates the extent to which  $\text{VO}_2\text{max}$ ,  $\% \text{VO}_2\text{max}$ , and LT are able to influence running performance (148). The ability to sustain a higher relative velocity while utilizing a lower  $\% \text{VO}_2\text{max}$  and avoiding LT equates to improved running performance. In this simple concept lies the power of RE.

Among the physiological and biomechanical factors that determine RE, stiffness, defined as a resistance to deformation when force is applied to an object, has been identified as a primary factor (27,46,98,128,132,146,147). Lower extremity stiffness, in the context of running, has been assessed on multiple levels from the global, the vertical excursion of the center of mass, to the local, being musculotendinous, joint, and passive stiffness. A growing body of literature has illustrated the effects of a multitude of interventions targeted at improving RE in both trained and untrained populations (84,96,109,111,112,128,131,132,156,159). Stiffness is the key determining factor in the utilization of elastic energy while running and has been positively correlated with superior RE (2,27,32,44,47,67,81,90,101). The majority of research regarding the effect of stiffness on RE has been focused on joint stiffness of the ankle and knee, and the musculotendinous stiffness of the shank musculature, specifically the triceps surae, all of which is assessed almost exclusively in the sagittal plane.

Recent investigations of fascia have demonstrated its capacity for transmitting force, resisting traction, adaptation to long-term stretch, and recoiling (77,78,102,135,160,161). These characteristics strongly suggest that the fascial structures of the lower extremity may play a pivotal role in the generation of passive forces and the utilization of elastic energy during running (142,143). Additionally, research of the elastic energy storage and utilization capacity of the iliotibial band (ITB) has opened new avenues of research into the capacity of previously ignored anatomical structures to aid in running economy (36,37). Collectively, these relatively new topics, along with the identification and exploration of fascial sling systems throughout the body represent unexplored realms of elastic energy utilization and running economy.

The purpose of this state-of-the-art review is to summarize the long-established conventions and more current research regarding the role of stiffness and elastic energy in running economy in order to chart aspects of the topic that require further investigation via primary research. Increased clarity on the role of stiffness in running economy will guide strength and conditioning coaches towards a better understanding of the potential lying dormant in distance runners and inform best practices on the development and exploitation of this potential. While the role stiffness plays in the utilization of elastic energy during running, and the impact this has upon running economy, has been established, the current understanding of the mechanisms governing it is limited by inaccurate methods of quantification and an abbreviation of the scope of investigation. Modern instrumentation, an ever-developing knowledge of the role of fascia in passive force generation, and a new understanding of the potential of the iliotibial band for elastic energy utilization opens new avenues for research. Further exploration of this aspect of exercise economy could provide invaluable insight into the generation and utilization of passive forces in cyclical human movement.

## 2. Methods

Literature included in this review was gathered from PubMed, Google Scholar, Semantic Scholar and Mendeley databases. Searches included studies, working papers, and reviews published between 2000 and 2021. Following initial screening of the search results forward and backward citations were also assessed which expanded the timeframe of included papers. Searches were conducted including combinations of the terms: “running economy”, “running mechanics”, “exercise economy”, “stiffness”, “fascia”, “fascial contractility”, “fascia lata”, “myofibroblast”, “musculotendinous stiffness”, “muscle-tendon unit stiffness”, “concurrent training”, “elastic energy”, “passive force”, “distance running performance”, “distance running”, “ground reaction forces”, “rate of force development”, “ground contact time”, “duty factor”, “metabolic cost”, “metabolic efficiency”, “running biomechanics”, “inverse dynamics”, “spring-mass model”, “joint stiffness”, “joint moment”, “joint kinetics”, “muscle-tendon elasticity”, “myotonometry”, “myotonometer”, “shear wave elastography”, “hysteresis”, and “iliotibial band”. Initial results (figure 1) revealed 1293 unique papers. Prior to screening 417 papers were removed from the pool due to redundancy or lack of an accurate translation to English. Of the remaining 876 papers 463 were discarded due to the lack of relevance to the act of running. Seven of the remaining 413 papers could not be retrieved for examination. The final 406 papers were filtered for inclusion in the review based upon: age of participants, health status of participants, and fitness status of participants. Forward and backward citations were evaluated based on relevance to running economy and/or stiffness and methodology. Relevant papers were included to establish context and scope to establish an accurate timeline for the development of relevant concepts, research methods and instrumentation.

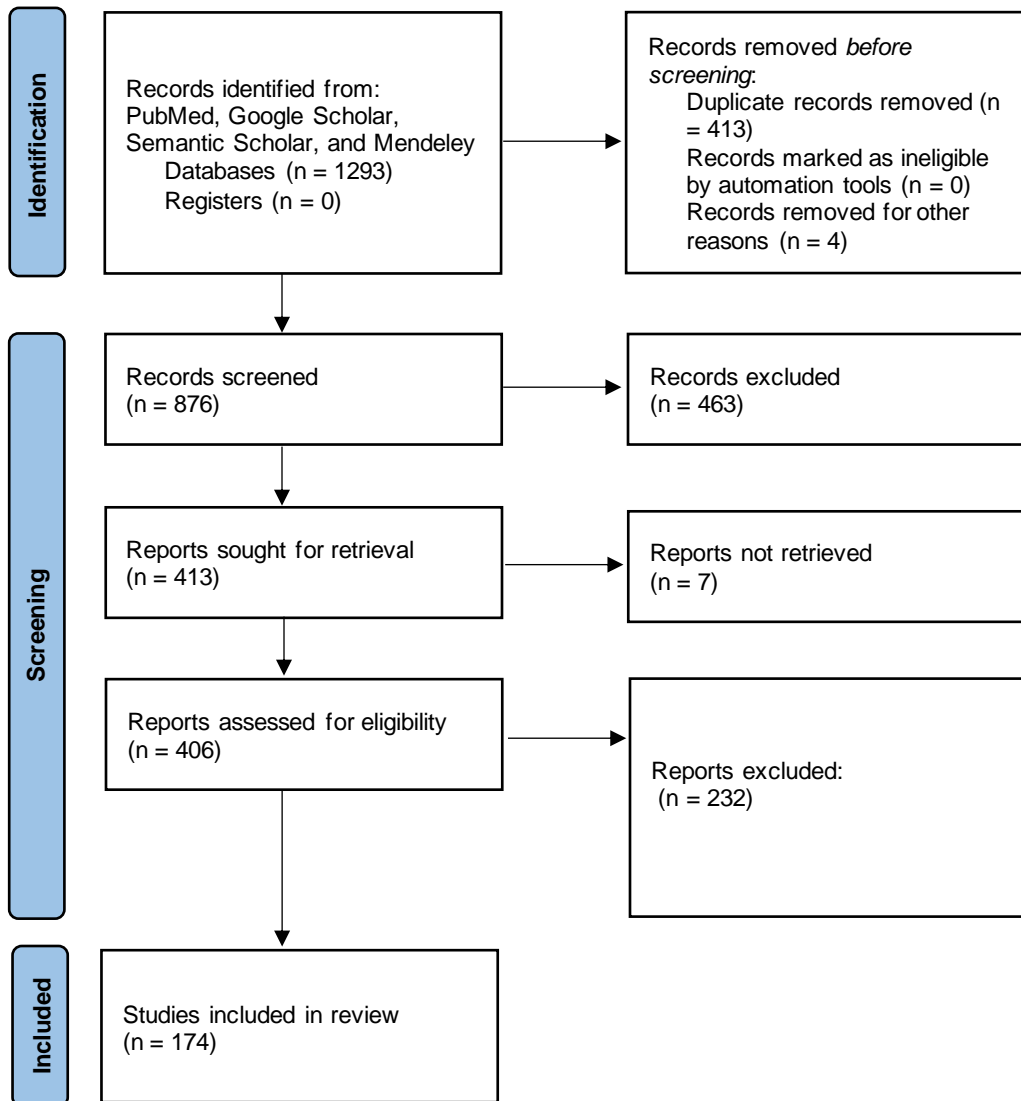


Figure 1. PRISMA flow chart.

### 3. Results

#### 3.1. Crucial Role of Running Economy

The history of distance running is littered with examples of athletes repeatedly dominating fields of their peers who possess superior  $VO_2\max$ , who, on paper, should themselves be dominant. This phenomenon was often chalked up to be the result of some combination of superior genetics, pain tolerance, superior lactate threshold, anatomical anomaly or superior training. Modern analysis of this observation has revealed RE may be the missing link. This

revelation has also demonstrated the significant extent to which RE dictates the influence that all other determinants of running performance have upon an individual's ability to perform in middle-distance to long-distance events (13,110,148). An individual's RE is determined by multiple physiological components, such as cardiovascular and neuromuscular systems, that influence biomechanical factors: force, power, ground reaction forces, stride frequency and length, and ground contact time (46,128,131,132). Collectively these factors determine the metabolic efficiency of an individual while running (46,128,131,132).

By definition,  $\text{VO}_2\text{max}$  is the maximum rate at which oxygen can be consumed by the body and utilized for exercise (12,13,71). Maximal oxygen uptake ( $\text{VO}_2\text{max}$ ) is considered the limiting factor for aerobic performance and is therefore considered the main determinant of performance in running events whose elapsed times exceed the capacity of anaerobic energy systems but fall short of the timeframe in which  $\% \text{VO}_2\text{max}$ , LT and RE influence performance. This relatively small window typically describes events ranging in distance from 1000 to 3000 meters. Literature devoted to the topic concludes that  $\text{VO}_2\text{max}$  is limited by the performance of the pulmonary system and circulatory system of an individual (13). The sustainable percentage of an individual's  $\text{VO}_2\text{max}$  for a given distance, known as the fractional utilization of  $\text{VO}_2\text{max}$  ( $\% \text{VO}_2\text{max}$ ), is a primary determinant of endurance performance. Whereas  $\text{VO}_2\text{max}$  effectively establishes a theoretical ceiling for performance potential,  $\% \text{VO}_2\text{max}$  and RE determine the percentage of that potential realized by an individual over a given distance. Both  $\% \text{VO}_2\text{max}$  and RE become increasingly important for running performance as distances, and therefore elapsed times, increase. Lactate threshold and  $\% \text{VO}_2\text{max}$  are often closely related. Lactate threshold (LT) is defined as the point during exercise at which blood lactate accumulates at a rate exceeding the rate at which it can be removed (13). Though  $\text{VO}_2\text{max}$ ,  $\% \text{VO}_2\text{max}$ , and LT

collectively dictate the upper limit of sustainable submaximal running performance, these parameters fail to describe the sustainability of said effort over a given distance. Running economy quantifies this final parameter of distance running performance by defining the metabolic cost of submaximal running performance. Multiple methods exist for the measurement of RE and despite variations in methods, protocol, and analysis, a consensus culminating in representative values has been established allowing for a degree of accuracy in the comparison of individuals given established parameters (46).

### **3.2. Determinants of Running Economy**

Investigations of RE have revealed several physiological components and biomechanical factors that determine an individual's RE. The physiological components that may have significant impact on RE include blood lactate concentration (LT), substrate metabolism, body core temperature, heart rate, and minute ventilation (8,98,123,132,148). The consensus of literature regarding the subject generally agrees that body core temperature, minute ventilation, and heart rate have a measurable but insignificant effect on exercise economy. The change in  $\text{VO}_2$  demand due to variation of these measures is too small to be considered a major contributing factor (8,124,134,139,166). Theoretically, an increased capacity for lactate buffering allows for a higher velocity at LT. Increased lactate buffering would directly influence  $\% \text{VO}_{2\text{max}}$ , narrowing the gap between  $\text{VO}_{2\text{max}}$  and  $\% \text{VO}_{2\text{max}}$ . Increased functionality of skeletal muscle mitochondria, increased fat metabolism, and a lower dependence on muscle glycogen serve to lower the metabolic cost of prolonged exercise and have been demonstrated to contribute significantly to improved RE (11,42).



Research has identified multiple biomechanical factors as primary contributors to superior RE. Stride frequency, stride length, ground contact time (GCT), swing time, and select ground reaction force (GRF) parameters, specifically rate of force development (RFD) and horizontal GRF, are chief among these (42,43,45,46). Horizontal impulse and peak horizontal force account for 33% of the metabolic cost of running (24). Anthropometric variables have also been assessed in RE research and revealed important relationships that influence RE. Of these, lower extremity mass distribution and Achilles tendon moment arm length have been shown to be the most significant (11,46,79,93,98,154,172). Research analyzing the biomechanical properties of East African distance runners correlated a tendency of mass distributed proximally in the lower extremities to increased RE (46,93,98). These findings have been supported by multiple studies examining the effects of distal loading of the lower extremities (104,127,136). Investigations of Achilles tendon moment arm length have revealed very strong correlations between short Achilles tendon moment arm and greater RE (11,79,154). This specific aspect of anthropometry, which is also directly related to joint stiffness, has a direct impact on musculotendinous stiffness by effectively increasing stiffness, promoting the utilization of elastic energy and limiting hysteresis resulting in improved RE. Gait characteristics have also been thoroughly explored in RE research. This particular body of literature strongly suggests that RE is closely associated with self-selected stride length and frequency (8,21,85,173) that generate the least amount of vertical oscillation (63,68,165,173) and therefore generates less vertical GRF and more horizontal GRF per stride (24,68,128,130).

### **3.3. Stiffness**

The exploration of RE, whether from the physiological or biomechanical perspective, inevitably dredges up questions surrounding underlying mechanisms. Though several of the

potential avenues of investigation remain to be thoroughly explored, one of the most significant underlying mechanisms, stiffness, in all of its forms, has revealed itself as the most potent neuromuscular component of RE. An abundance of experimental literature has identified stiffness as a potential mechanism driving several other variables linked to superior RE, both neuromuscular and biomechanical, including RFD and GCT (46,101,128,132,146,147). The consensus of literature examining the effect of stiffness upon running performance strongly suggests the relationship between stiffness and elastic energy utilization is the most significant determinant of RE (46,101,128,132,146,147).

In the context of exercise science, stiffness as a metric, is assessed on multiple levels including vertical, whole-leg, joint and musculotendinous stiffness (2,11,27). Strictly speaking the mechanical definition of stiffness does not apply to the complex human body due to the presence of a multitude of tissues all of which have unique biomechanical properties that do not resemble the simple passive bodies found in the classic definition of stiffness (174). Though the term ‘stiffness’ is still most often used in literature to refer to the biomechanical property of stiffness, researchers have also offered up the more suitable term quasi-stiffness to refer to the property and redefined the concept to better describe stiffness in biomechanics. Quasi-stiffness describes the ability of the human body, or referenced segment, to resist displacement by external forces (174). Quasi-stiffness only refers to instantaneous stiffness and does not consider the time course of the displacement (174). The biomechanical convention of stiffness is based upon a simple approximation of the action of an elastic body when force is applied to that elastic body known as Hooke’s Law. The resulting equation  $F=kx$  ( $F$  = force,  $k$  = stiffness,  $x$  = deformation) can be used to calculate stiffness when a known force is applied to an elastic body and the deformation of that body is measured. Because both stiffness and quasi-stiffness are used

interchangeably in biomechanical, and physiological literature the term ‘stiffness’ will be used in this review to refer to the biomechanical property of stiffness. The examination of stiffness has inspired the genesis of multiple models created with the intention of better explaining the biomechanical nuance of the human running gait. These approximations of the complex spring-like lower extremities have progressed considerably since the initial examinations of human running but, to date, no single model has allowed for an accurate and comprehensive examination.

### **3.4. Measuring Stiffness**

#### *Modeling the Lower Extremity*

The spring-mass model, one of the earliest attempts at modeling the lower extremities, long persevered as the most commonly utilized approximation with the adaptation of more complex methods only recently becoming more common. This simple model describes a mass, representative of a subject’s whole-body center of mass (COM), resting above a weightless spring representing the lower extremity (27). During locomotion, forces in the form of GRF compress the spring during the eccentric initial phase of ground contact, storing mechanical energy in the spring, which is subsequently released as the spring recoils during the propulsive phase of ground contact (27,106,125). The degree to which the lower extremity resists compression (deformation) is dependent upon the stiffness of that element, in this case whole-leg stiffness, which is dependent upon the combination of muscle, tendon and aponeurosis stiffness (2,5,19,120). The amount of elastic energy return, or utilization, during the propulsive phase of ground contact is positively correlated to stiffness and ultimately RE (3,5,27). This concept of whole-leg stiffness is compounded by the inclusion of three main joints, the ankle, knee, and hip.

These joints effectively divide the whole-leg element into three individual segments, which complicates the assessment of stiffness further as the stiffness of the joints must also be considered. The simplicity of the spring-mass model provides researchers with a straightforward method for quantifying the vertical stiffness of an individual but fails to provide the detail necessary to identify the underlying mechanisms or assess the stiffness of individual elements in the kinetic chain. Additionally, the simple spring-mass model is limited to two dimensional assessment.

Inverse dynamics has been used extensively to analyze the kinematics and kinetics of running and is the basis upon which modern computer-generated modeling is built (39,91,138,157,168). Inverse dynamics, a technique which allows for the calculation of multibody dynamics, utilizes the Newton-Euler equations for force (*force = mass x linear acceleration;  $F = ma$* ) and moment (*moment = mass moment of inertia x angular acceleration;  $M = Ia$* ) and theoretical link-segment model of the lower extremity to calculate the joint moments at the ankle, knee, and hip to derive the kinetics of the leg (14). Though inverse dynamics has proven itself as a valuable tool, and is therefore widely used, the technique is sensitive to the accuracy of the instrumentation and skill of the technicians conducting the data collection and is based upon several assumptions that challenge the accuracy of results (75,76,105). Of the assumptions inherent to inverse dynamics, several present direct conflicts with the accuracy of this technique to describe the kinetics of the leg during running. These include: 1) the joints are frictionless pin-joints 2) the segments of the leg are rigid 3) the mass of each segment is concentrated at the calculated center of mass, and 4) there is no co-contraction of agonist and antagonist muscles. A basic understanding of anatomy coupled with a wealth of research clearly illustrates that these assumptions are obviously not representative of reality and challenge the

accuracy of an inverse dynamics analysis of the human leg during running (18,39,59,91). The limitations of the inverse dynamics method result in the inaccurate calculation of joint reaction forces, joint stiffness and work performed at the joint. Additionally, the method fails to account for the different biomechanical characteristics of all the tissues present in the lower extremity. This represents a significant barrier to the accurate assessment of the underlying mechanisms of RE, specifically regarding the role of the hip joint and associated bi-articular MTUs, and the influence of fascia and fascial sling systems.

It has been suggested that the central error found in the inverse dynamics method lies in the incompatibility of data gathered from the organic subject, which behaves in a very specific and yet to be fully understood manner, being used as inputs for a mathematical model which behaves in an unrealistic manner (65). The Hill muscle models, upon which the majority of the musculoskeletal models utilized by simulation software incorporating inverse dynamics are based, have been found to be inaccurate in their simulation of the behavior of muscle tendon units (MTU) during running (58,155). Specifically, it has been demonstrated that these models lack sensitivity in their description of the behavior of the parallel elastic element and the force velocity curve of the contractile element during the gait cycle (58,155). Furthermore, all commonly used muscle models ignore the influence of fascia and fascial sling systems which further complicates the inaccuracies. Classically, fascia and aponeuroses have been falsely represented or failed to be represented altogether in many musculoskeletal models. This has resulted in inaccurate calculation of the stiffness of series elastic elements, and misrepresentation of the stiffness and Young's modulus (ratio of tensile stress to tensile strain of a material) of aponeuroses and fascia (70).

Several attempts have been made to improve the accuracy of inverse dynamics via optimization algorithms (39), smoothing algorithms (59) and least-squares estimation computation (18,91). These attempts at improving the precision of inverse dynamics analysis provide valuable insight into the nature of the error present in the technique. Classic inverse dynamics calculation starts at the most distal segment (foot contact with a force plate) and proceeds proximally to calculate all successive parameters. Continual error begins to compound and creates substantial error in the calculation of all variables and the accumulation of residual forces and torques (18,91). Due to this phenomenon, joint reaction forces, joint stiffness, and work performed at the joint are often inaccurately estimated. Evaluation of the hip joint, in particular, is problematic and is therefore often not assessed as a variable in running studies.

### *Vertical Stiffness*

Vertical stiffness is a measurement of the vertical excursion of the COM during the stance phase of running. This quantification of the global stiffness of a subject is the broadest value utilized in the assessment of stiffness during locomotion and remains one of the most commonly assessed types of stiffness in the literature (1,3,55,57,66,72,80,90,99,100,115,116,16,117,125,140,145,21,27,34,41,52–54). The calculation of vertical stiffness has evolved as instrumentation has become more advanced and allowed for more in-depth and externally valid measurements of kinetics possible. McMahon et al. formulated the initial method to calculate vertical stiffness proposing that vertical stiffness ( $k_{vert}$ ) was equal to the product of mass and the square of the natural frequency of oscillation: ( $k_{vert}=m\omega^2$ ) (107). McMahon derived the natural frequency of oscillation from contact time and vertical velocity which was calculated from a vertical force and contact time curve created by data gathered from the utilization of a force plate (107). McMahon's initial method was

improved upon in the work by Cavagna et al. in which a natural frequency of oscillation ( $\omega$ ) was again derived from solely ground (force plate) contact time from which a period of oscillation ( $P$ ) was calculated (22). This resulted in the formula:

$$k_{\text{vert}} = m \left( \frac{2\pi}{P} \right)^2$$

where  $k_{\text{vert}}$  = vertical stiffness,  $m$  = mass, and  $P$  = period of oscillation (22). McMahon and Cheng, in an attempt to create a simplified model for running in order to examine the relationship between velocity and stiffness, calculated stiffness from maximum vertical force and maximum vertical displacement expressed as:

$$K_{\text{vert}} = \frac{F_{z \text{ max}}}{\Delta y}$$

where  $K_{\text{vert}}$  = vertical stiffness,  $F_{z \text{ max}}$  = maximum vertical ground reaction force, and  $\Delta y$  = vertical displacement of the subject's center of mass during ground contact (106). The resulting method states that vertical stiffness is equal to the change in vertical force divided by the change in vertical position ( $\Delta f_y / \Delta y$ ) during the mid-stance phase of running (106). In an attempt to simplify the popular model created by McMahon and Cheng, Morin et al. developed a technique that presents the possibility of vertical stiffness calculation beyond the confines of the laboratory (125). Morin's technique, an adaptation of a model used by Dalleau et al. (28), calculates vertical stiffness from mass, ground contact time, and aerial time, the latter two variables being measured via pressure sensors worn by the experiment participants in their shoes. This data is then used to create a sine wave-based force-time curve model from which vertical stiffness can be calculated (125). This model considers force as a simple sine function expressed as:

$$F(t) = F_{\text{max}} \sin\left(\frac{\pi}{t_c} t\right)$$

with  $F_{max}$  = peak force and  $t_c$  = contact time (125).

Using this equation, vertical stiffness ( $\hat{k}_{vert}$ ) was then calculated as the ratio of the maximal force ( $\hat{F}_{max}$ ) over the vertical COM displacement ( $\Delta\hat{y}_c$ ) expressed as:

$$\hat{k}_{vert} = \hat{F}_{max} * \Delta\hat{y}_c^{-1}; \text{ with } \hat{F}_{max} = mg \frac{\pi}{2} \left( \frac{t_f}{t_c} + 1 \right)$$

with  $m$  = mass,  $t_f$  = flight time,  $t_c$  = contact time

$$\text{and } \Delta\hat{y}_c = \frac{\hat{F}_{max} t_c^2}{m\pi^2} + g \frac{t_c^2}{8} \quad (125).$$

To verify the validity of this novel technique Morin et al. compared data derived from the sine wave-based method to contemporary force plate derived stiffness data and concluded that the new technique offered an acceptably accurate approximation (125). Currently, the calculation of vertical stiffness is most often accomplished via data collection with a multi-camera motion capture system in conjunction with a force plate(s) and calculation performed by proprietary software. This software often employs a classic inverse dynamics calculation but a simple calculation of vertical stiffness can be accomplished using McMahon and Cheng's method in conjunction with a force plate and simplified marker placement schedule (in order to orient a subject's COM). This more advanced application of McMahon and Cheng's original method has eliminated much of the tendency for overestimation of the displacement of COM that is the common error witnessed with the original technique and therefore provides a significantly more accurate measurement of vertical stiffness (3).



## *Whole-leg Stiffness*

Moving from the global measurement of vertical stiffness towards more specific measurements of lower extremity stiffness is crucial in the quest for greater understanding of the underlying mechanisms of stiffness in order to better understand how these mechanisms can be manipulated. Whole-leg stiffness, also frequently referred to as leg stiffness or leg spring stiffness, is the measurement of the stiffness of the whole leg when the three-segment limb is considered as a single spring-like structure. Due to the relative ease of measurement, whole-leg stiffness, much like vertical stiffness, is very commonly assessed in contemporary literature despite the availability of more granular metrics (1,3,57,66,72,80,97,99,100,115–117,7,125,126,140,145,158,170,16,21,34,41,52–54). McMahon and Cheng's popular method for calculating vertical stiffness also resulted in a prevalent method for calculating whole-leg stiffness (106). The method utilized the vertical force data and COM vertical excursion measurement described above with measurement of velocity, instantaneous leg segment length, and whole-leg angle at the moment of initial impact to calculate change in whole-leg length using the following formula:

$$\Delta L = \Delta y + L_0(1 - \cos \theta)$$

where  $\Delta L$  = length change of the leg-spring,  $\Delta y$  = vertical excursion of COM,  $L_0$  = baseline COM position, and  $\theta$  = angle of leg-spring model in reference to vertical axis at initial contact (106). Using these data leg stiffness was then expressed as a ratio of maximum vertical force to maximum change in leg length (106). Morin et al. (125) also applied their previously mentioned simplified method for modeling running to the measurement of whole-leg stiffness. Foregoing

the force plate, Morin et al. (125) used running velocity, measured with a radar gun, and initial leg length to calculate whole-leg stiffness from the maximum change in leg length.

$$\hat{k}_{\text{leg}} = \hat{F}_{\text{max}} * \Delta\hat{L}^{-1}$$

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

and  $\hat{k}_{\text{leg}}$  = leg stiffness,  $\hat{F}_{\text{max}}$  = maximal force,  $\Delta\hat{L}$  = change in leg length,  $L$  = leg length,  $\Delta\hat{y}_c$  = COM displacement,  $v$  = velocity, and  $t_c$  = contact time (125). Arampatzis et al. (3) conducted a study to compare the mathematically calculated leg length changes with those measured kinematically. Arampatzis and colleagues employed a force plate (sampling at 1000 Hz) and two high-speed cameras (sampling at 120 Hz) and a system of reflective markers to conduct data collection. Following this, whole-leg and vertical stiffness were calculated using formulas already established previously in research (106,107):

$$K_{\text{leg}} = \frac{F_{z \text{ max}}}{\Delta L} \text{ and } K_{\text{vert}} = \frac{F_{z \text{ max}}}{\Delta y}$$

with  $K_{\text{leg}}$  = leg stiffness,  $F_{z \text{ max}}$  = maximum vertical ground reaction force,  $\Delta L$  = leg length change,  $K_{\text{vert}}$  = vertical stiffness, and  $\Delta y$  = vertical displacement of the subject's center of mass (3,106). The leg stiffness values reported by Arampatzis and colleagues (3) far exceeded those calculated by any previous study employing the established mathematical stiffness calculation methods. Ultimately the researchers argue that this difference exists due to a tendency for overestimation of the excursion of the center of mass that is inherent when using the method made popular by McMahon and Cheng (106).

## *Joint Stiffness*

Moving further in the direction of granularity, joint stiffness becomes key in the discussion of specific contributions to elastic energy utilization. Joint stiffness, often described as a torsional spring, is the ratio of joint moment to the angular displacement of that joint and thus falls outside of the strict definition of stiffness. Because joint work and power are directly related to running velocity, a thorough understanding of joint stiffness is necessary in understanding the role of joint stiffness in the limitation of hysteresis and utilization of elastic energy associated with superior RE (81,164). The creation and implementation of a torsional spring model of leg/“leg-spring” stiffness has allowed researchers studying lower extremity stiffness to identify the contribution of each joint (ankle, knee, and hip) to overall leg stiffness (3,61,164,167,170,64,81,90,103,108,158,162,163). This has benefitted research on the topic by aiding in the assessment of the relationship between velocity, stiffness and RE (discussed in-depth later), and, along with EMG, demonstrated how joint stiffness is modulated in each joint as velocity changes. The torsional spring model of leg stiffness can be described by the equation:

$$K_{\text{leg-spring}} = \frac{1}{(1/K_{\text{ankle}}) + (1/K_{\text{knee}}) + (1/K_{\text{hip}})}$$

which dictates that the greatest influence over overall leg stiffness is the most compliant or least-stiff joint at any instance during ground contact (3). Commonly accepted methods employed for the measurement of joint stiffness utilize the same data collection methods as contemporary studies of whole-leg and vertical stiffness (ground reaction force data from force plate(s) and digitized kinematic data from high-speed video footage in conjunction with a marker system indicating anatomical landmarks) with the additional incorporation of inverse dynamics computations to calculate joint moments (61,81,90,162). Given this data, joint stiffness is

calculated as the change in joint moment ( $\Delta M$ ) divided by the change in joint angle ( $\Delta\theta$ ) (61,90,162):

$$K_{\text{joint}} = \frac{\Delta M}{\Delta\theta}$$

Typically, these inverse dynamics calculations are performed via the same software responsible for recording and processing the force plate and kinematic data. An exception is represented in the work of Arampatzis et al (3) who calculated joint stiffness along with leg stiffness in order to accurately evaluate the relationship between velocity and stiffness in human running. The researchers used the same kinematic data gathered for the other elements of the study and modeled the ankle joint and knee joint as “rotational springs”. The calculation was expressed as a ratio of negative work to change in joint angle: ankle stiffness ( $K_{\text{ankle}}$ ) is equal to two times the negative mechanical work of the ankle joint ( $W_{\text{ankle}}$ ) divided by the change in ankle angle ( $\Delta\theta_A$ ); knee stiffness ( $K_{\text{knee}}$ ) is equal to two times the negative mechanical work of the knee joint ( $W_{\text{knee}}$ ) divided by the change in knee angle ( $\Delta\theta_K$ ):

$$K_{\text{ankle}} = \frac{2W_{\text{-ankle}}}{\Delta\theta_A^2} \quad \text{and} \quad K_{\text{knee}} = \frac{2W_{\text{-knee}}}{\Delta\theta_K^2}$$

The researchers calculated the joint moments and power using the inverse dynamics methods outlined by Hof (73). This method later received criticism on multiple levels, which claimed it was irreproducible and has not been utilized since (61). All other studies employ standard inverse dynamic calculation methods or modelling software which calculates joint stiffness, among other variables, using inverse dynamics (61,81,90,162).

## *Muscle-Tendon Unit Stiffness*

Considering the stiffness of individual muscle-tendon units (MTU) allows for the analysis of stiffness within each segment of the multi-segment lower extremity.

Musculotendinous stiffness is a key contributor to whole-leg stiffness and is positively correlated to RE (2,4,19,106,118,120). The most common technique for the measurement of MTU stiffness is the free oscillation technique (31,168,169), though alternative methods exist for the measurement of tendon stiffness (94) and new methods continue to be explored and validated (23,88,133). The free oscillation technique models the MTU system, consisting of the contractile element (CE), series elastic element (SE), and parallel elastic element (PE), as a damped mass-spring system (31). Free oscillation is most commonly used in the assessment of triceps surae, knee flexors, and knee extensors. The technique involves loading a MTU in a given body segment and then perturbing that segment which results in damped oscillations which are measured and used to calculate stiffness (31). To elaborate, the subject being evaluated has the MTU in question loaded with successively heavier weights within a range (ex. 5-40% (33)) of their established maximal isometric contraction of that MTU. Each of these loads are then perturbed and the resulting oscillation of the MTU is measured using a force plate (33). The oscillations of the MTU system can be modeled by the motion equation:

$$F = -kx - hv$$

where  $F$  = force,  $k$  = stiffness of the MTU spring model,  $x$  = the displacement,  $h$  = the viscous element (dampening force), and  $v$  = velocity (31). It is important to note, in this model, the measure of stiffness ( $k$ ) most accurately corresponds to the SE and PE (and similar elastic tissue and bone), while the viscous element, which acts as the harmonic dampener, most closely

corresponds to the CE (31). When this simple equation is developed with substitutions for the first and second derivatives of displacement a repetitive function expressing angular natural frequency emerges from which harmonic motion can be calculated (31). Given this, it becomes clear that angular natural frequency is proportional to the modeled MTU stiffness and inversely proportional to the mass of the MTU (31).

Tendon stiffness, while often included in the assessment of MTU stiffness, has been measured in isolation and has also been positively correlated to RE (43). The most widespread method for the assessment of tendon stiffness is ultrasonography (85,86,89) though more advanced techniques have also shown promising degrees of validity (24,94). The application of ultrasonography to the measurement of tendon stiffness employs the standard stiffness formula ( $F = kx$ ) but derives force ( $F$ ) from joint torque, and displacement ( $x$ ) from tendon excursion which is measured using ultrasonography (43,85,86,89). In order to do this, the subject being assessed is secured to a dynamometer (typically by either the foot or shank depending upon the tendon being measured) and the segment is moved through a preprescribed range of flexion and extension. Resistance by the MTU(s) during the procedure is recorded as passive joint torque. An established method for the conversion of the measured torque to muscle force uses the formula:

$$F_m = k * T * d^{-1}$$

where  $F_m$  = muscle force,  $k$  = relative contribution of the cross-sectional area of the CE of the MTU being assessed,  $T$  = measured torque from the dynamometer procedure, and  $d$  = moment arm of the MTU being assessed (86,87). During the dynamometer procedure, real-time ultrasound is utilized to measure the elongation of the tendon being evaluated. Accurate assessment of the tendon excursion is accomplished by identifying the point at which a single

fascicle attaches to the aponeurosis of the tendon being measured while at rest, and then measuring the maximal displacement of that attachment point during the dynamometer procedure (86,87).

Passive stiffness of MTU and tendon have been evaluated using the previously discussed isokinetic dynamometer and free oscillation methods, but also via shear wave elastography and myotonometry (23,94,114). Shear wave elastography, while being utilized extensively across many disciplines, has also been used specifically to illustrate the relationship between trends in passive MTU stiffness and distance running performance (114). Shear wave elastography measures passive stiffness by generating an acoustic radiation force in the form of a focused pulse from a hand-held unit and then measures the speed of the propagation of the shear wave as it reacts to the stiffness of the tissue upon which the beam was focused. The shear wave speed is then interpreted as a representative of tissue tensile strength because of its correlation to elastic moduli (94,114).

Myotonometry is a relatively new development. Early attempts at developing accurate myotonometers with acceptable levels of repeatability were problematic but a new generation of myotonometers are capable of valid robust assessment when compared to contemporary methods. Myotonometry measures passive stiffness in a similar manner to the free oscillation technique in that the calculation of stiffness is derived from a measurement of damped oscillation but differs in its method of causing the oscillation. The most recent development in myotonometry is a hand-held digital palpation device that generates a mechanical force into a targeted tissue and measures the resultant damped oscillations with a built-in accelerometer and calculates stiffness as the ratio of resistance to external force and the maximum displacement of the target tissue (23,94). Though the utility of myotonometry is yet to be fully explored, the

presence of accurate and valid methods of assessing passive stiffness *in vivo* allows for the development of more accurate musculoskeletal modelling, and the assessment of acute and long-term training effects. Myotonometry may also be a key investigative tool in the exploration of the role fascia, and fascial sling systems, play in exercise economy.

### **3.5. Relationship Between Stiffness and Running Economy**

The collective assessment of the pertinent literature strongly suggests that stiffness is the most significant determinant of RE due to its role in the successful utilization of elastic energy during running (46,101,128,132,146,147). The primary observation involves the relationship between GCT and RE, and the insight this observation has had upon subsequent investigations of stiffness. Studies exploring GCT as a variable or determinant of RE also frequently assess RFD and duty factor (DF) as parallel lines of examination. Many of the revelations specific to the investigations of anthropometry and running mechanics in East African distance runners have generated the questions surrounding the underlying mechanisms of RE that drive the research in its current direction.

Runners from East African nations including Uganda, Eritrea, Kenya, Ethiopia and Somalia have dominated the middle and long distance running world for several decades. Given this phenomena it is not surprising that a large body of research has been dedicated to evaluating the determining factors behind this success (93,121,122,146,147). Running economy was identified early in these investigations as a primary determinant of running performance and quickly inspired additional research aimed at identifying the underlying mechanisms (93). Though many questions still remain regarding the inherent mechanisms of East African dominance, the common theme of short GCT being correlated to superior RE, has led to further



investigation of GCT and MTU, joint, and aponeurosis stiffness (17,74,121,122,144).

Subsequent investigation demonstrated that increased stiffness of MTUs, joints and aponeurosis has been negatively correlated with the energetic cost of running (27,111,131,132,159).

Greater musculotendinous stiffness is positively correlated with improved RE via improved elastic energy utilization and decreased GCT (27). The relationship between GCT and stiffness is indicative of an improvement in the efficiency of the stretch-shortening cycle (SSC). Stretch-shortening cycles occur when MTUs are actively lengthened and then concentrically contracted as in running or a counter movement jump (27). As the MTU is stretched, in the case of running via the eccentric muscle action occurring during the initial phases of ground contact, elastic strain energy is stored in the connective tissue. Upon concentric contraction, during the later phase of ground contact, this elastic strain energy can add to the force and power generated by the contraction as the elastic components of the involved MTUs recoil (47). This 'recycled' energy significantly reduces the amount of force, and therefore energy, required to maintain the cyclical action of running. The amount of elastic energy that is used or lost during the SSC is determined by multiple physiological and biomechanical factors and is analogous to the phenomena of hysteresis (47). It follows that a reduction in the hysteresis, or parasitic loss of energy, in a SSC would greatly benefit the utilization of elastic energy (27).

An increase in leg stiffness was suggested as a key mechanism in an improvement in RE in a study conducted by Paavolainen et. al, due to the significant decrease in GCT in conjunction with the significant increase in peak sprint velocity, and the concurrent improvement in jump testing (132). Paavolainen and colleagues examined concurrent training effects on running performance and RE. They examined GCT as a dependent variable in order to explore the mechanisms of neuromuscular adaptation resulting from the training program (132). The

investigators found significant reduction in GCT with no significant concurrent alteration of GRFs (132). This shortening of GCT was suggested to be a significant contributing factor in the significant improvements in running performance and RE also observed amongst the experimental group in the training study (132). The findings of this study were later supported by further inquiry performed by Paavolainen et al. and Nummela et al. (128,131). Both studies investigated neuromuscular characteristics, muscular power, and running performance in order to examine the interaction between running biomechanics, velocity, and RE. Paavolainen et al. reported significant correlations between RE and GCT during a 5 km time trial ( $r=0.64$ ). Nummela et al. reported significant correlations between RE and GCT at three of the four velocities tested:  $5.8 \text{ m}\cdot\text{s}^{-1}$  ( $r=0.49$ ),  $6.2 \text{ m}\cdot\text{s}^{-1}$  ( $r=0.44$ ) and  $6.6 \text{ m}\cdot\text{s}^{-1}$  ( $r=0.41$ ). Additionally, both studies found a direct negative correlation between GCT and velocity ( $r=-0.52$  (128);  $r=-0.49$  (131)) during a running time trial, which also correlated significantly with RE ( $r=-0.47$  (128);  $r=0.64$  (131)). These findings are also reflected in the work of Tam et al. (164) who reported a significant correlation between reduced GCT and RE ( $r=0.639$ ) as well as several other studies that have sought to identify the biomechanical characteristics of runners with superior RE (11,25,98,106).

Duty factor (DF) is defined as the ratio of GCT to the duration of the aerial portion of the running stride commonly referred to as the swing phase. The greater the GCT the greater the DF. An analysis of DF aids in the examination of the interaction between leg stiffness, elastic energy utilization, and RE (101). The vast majority of research investigating RE that accounts for GCT and any aspect of stiffness strongly suggests that RE is negatively correlated to GCT and positively correlated to stiffness, both of which imply a low DF (11,25,98,128,131,164). This consensus is challenged by the recent findings of Lussiana et al. (101). The researchers collected

kinematic and kinetic data on fifty-four trained runners organized into two groups identified as either low- or high-DF. Data was collected at 10, 12, 14, 16 and 18 km·h<sup>-1</sup> and showed no significant effect of DF on RE ( $p = 0.556$ ) but did report an effect of speed on RE that was dependent upon DF ( $p = 0.025$ ). Specifically, the study revealed a significant decrease in RE ( $-2.3 \pm 2.6\%$  from 10 to 14 km·h<sup>-1</sup>,  $p = 0.008$ ) for the low-DF group but no significant change for the high-DF group ( $1.5 \pm 3.8\%$  from 10 to 14 km·h<sup>-1</sup>,  $p = 0.781$ ). Interestingly the researchers also reported an increase in step symmetry (ratio of concentric phase of ground contact to eccentric phase of ground contact) with an increase in velocity. This relationship between velocity and step symmetry strongly suggests that the mechanism underlying the observed interaction between velocity and DF, and therefore GCT, is directly related to a nearly isometric action of the muscles of the leg at higher velocities. A near-isometric action of the leg muscles (contractile element), which effectively increases the muscle stiffness, would promote greater energy storage and release from the elastic components of the leg including both series elastic components and parallel elastic components (44,101).

The observations regarding the relationship between DF and velocity, and the implications these observations have upon the relationship between stiffness and RE, are echoed in many other studies examining the mechanisms governing RE (33,81,95,129,131,164). Multiple studies have demonstrated a very strong interaction between stiffness, velocity, and RE. Li et al. measured whole-leg stiffness at three speeds (12, 14, and 16 km·h<sup>-1</sup>) and found a significant correlation between RE and velocity (95). The researchers observed a linear relationship between stiffness and velocity. As velocity increased, the correlation between whole-leg stiffness and velocity also increased ( $r = -0.686$ , 12km·h<sup>-1</sup>,  $p < 0.001$  and  $r = -0.761$ , 16 km·h<sup>-1</sup>,  $p < 0.001$ ). Dumke et al. focused on the MTU stiffness of the triceps surae, analyzing

the muscle and tendon separately and noted a significant negative correlation ( $r = -0.69$ ,  $p = 0.01$ ) between muscle stiffness at high velocities ( $322 \text{ m} \cdot \text{min}^{-1}$ ;  $19.32 \text{ km} \cdot \text{h}^{-1}$ ) and  $\text{VO}_2$  (improved RE)(33). Interestingly, Dumke and colleagues reported no significant correlations between muscle stiffness and  $\text{VO}_2$  at slower velocities or tendon stiffness and  $\text{VO}_2$  at any measured velocity. The researchers noted that this difference in their findings, compared to those of similar studies, was likely due to the inherent accuracy issues with the measurement method employed (free oscillation) and the restrictions of obtaining stiffness values of an MTU at a single joint angle (33).

More contradictory results were reported by Giovanelli and colleagues who assessed vertical stiffness and whole-leg stiffness in ultra-marathoners. The investigators reported an increase in vertical stiffness ( $+7.2 \pm 1.0\%$ ) but a decrease in leg stiffness as a function of speed ( $-4.6 \pm 5.1\%$ ) (56). The important take away arose from the investigators note that the submaximal speeds at which the subjects were tested were not consistent with typical speeds sustained during a race, and therefore may not have elicited the same neuromuscular response in the form of musculotendinous stiffness (56). This suggestion is reminiscent of the findings of multiple studies that demonstrate a relationship between self-selected pace, optimal vertical/leg-stiffness and superior RE.

More typical findings, which tend to employ the most current instrumentation, and higher testing velocities, have been reported from studies examining joint work, power and stiffness. Jin et al. analyzed joint work, power and stiffness at multiple velocities ( $1.8\text{-}3.8 \text{ m/s}$ ) and found significant correlations between velocity and joint stiffness at the ankle, knee and hip ( $r=0.96$  ankle,  $r=0.98$  knee,  $r=0.82$  hip,  $p < 0.05$ )(81). Tam et al. (164) measured several biomechanical and neuromuscular factors in trained distance runners in order to identify characteristics

indicative of more economical distance runners. Among other variables, joint stiffness ( $\text{Nm} \cdot \text{deg}^{-1}$ ) was assessed and compared to the subject's oxygen consumption ( $\text{ml} \cdot \text{kg}^{-1} \cdot \text{km}^{-1}$ ) to evaluate its contribution to RE. The experiment revealed a significant positive correlation between ankle stiffness and oxygen consumption ( $r=0.527, p=0.007$ ) and a significant negative correlation between knee stiffness and oxygen consumption ( $r=0.384, p=0.043$ ). These findings are supported throughout the literature though most similar studies note the positive correlation between ankle stiffness and oxygen consumption is only present during constant velocity running (3,40,61,90,113,162,163,170).

A generous body of research has been dedicated to training studies aimed at improving RE via a wide range of interventions applied to distance runners across a broad spectrum of fitness, age, and experience. A large portion of these investigations have been targeted at resistance training interventions, many of which have been instrumental in illustrating the important connection between increased vertical, whole-leg or joint stiffness and improved RE (47,60,132,137,156,159,62,84,95,96,109,111,112,131). Consequently, multiple systematic reviews and meta-analyses have been published to determine if a single training modality is superior in its training effect upon running performance or RE. The findings of these reviews vary slightly but generally agree that resistance training is beneficial and one of the primary beneficial outcomes of resistance training is increased stiffness (10,20). Specifically, while some interventions are more effective than others, interventions that improve the neuromuscular characteristics of the lower limb in regards to pre-activation of the agonist and antagonist MTUs, and MTU and joint stiffness have the greatest beneficial effect on RE (10,20). The important deduction stemming from this research is the strong evidence that stiffness is a trainable characteristic and is demonstrably correlated to improved RE.

Of particular interest are the studies incorporating concurrent training (CT), the combination of typical running training with resistance training or plyometric training, interventions with the assessment of stiffness and RE. Albracht et al. conducted such a study and demonstrated a significant increase in tendon-aponeurosis stiffness of the triceps surae (15.8%,  $p < 0.01$ ) in conjunction with an increase in RE (4.35%,  $p < 0.05$ ) resulting in significant group-by-time interactions with  $\text{VO}_2$  consumption ( $p = 0.03$ ) and energy cost ( $p = 0.02$ )(2). The experimental group showed a 4.2% ( $p < 0.05$ ) reduction in energy cost per body weight and a 4.8% ( $p < 0.05$ ) reduction in  $\text{VO}_2$  consumption while the control group showed no change (2).

While many studies have been conducted to address the effects of resistance training on running performance there still exists a void of knowledge specifically regarding the effect of training on mechanical stiffness. Stiffness and RE have, to date, primarily been assessed as dependent variables in the context of training studies investigating the effects of said training on running performance. These studies are very suggestive and allow for the speculation of relationships between the two variables but leave much to be desired. Considerable investigation is still required to assess the specific effects of popular training modalities on all forms of stiffness in the lower extremity. Only then can a clearer picture be formed of why resistance training alters RE and, consequently, a better understanding of how stiffness is involved in the utilization of elastic energy.

### **3.6. Role of Fascia in Elastic Energy Utilization**

Along with exploration of GCT, investigations of passive stiffness, co-activation of biarticular muscles, fascial stiffness, and fascial contractility have provided important data regarding the time course of force development. These insights have opened new lines of inquiry

regarding the role fascia and passive force transmission play in the utilization of elastic energy. Until recently, fascia was largely dismissed as an undifferentiated tissue with little function beyond the encapsulation of MTUs, organs and connective tissue. Interest in fascia and its function followed in the wake of advancements in technology and research methodology. Fascial dissection has demonstrated that fascia is a complex matrix of inelastic collagen fibers mixed with elastic fibers (161). Each unique layer of this woven matrix is separated by thin layers of adipocytes which function as lubricant allowing adjacent layers to freely slide over each other (160,161). This organization allows fascia to function in multiple, seemingly contradictory roles: resist traction, adapt to long-term stretch, recoil, and transmit force (135,160,161). The ability of fascia to transmit force has been demonstrated in animal studies, cadaver studies, and via modelling (77,78,102,135).

Current research strongly suggests fascia permeates the body with an innervated, adaptive connective tissue that functionally integrates, and aids in the coordination of movement. Multiple studies have demonstrated the presence of myofibroblasts in human fascia and the ability of this muscle tissue to contract when in the presence of contraction-inducing agents (149–153). These discoveries have interesting implications regarding the nature of elastic energy utilization. Schleip et al. (149) have demonstrated the ability of fascia to actively contract due to the presence of smooth muscle myofibroblasts. While the force generated by these contractions is calculated to be below the threshold for a positive contribution to joint stability/stiffness, it is strongly suggested that fascial contraction may positively affect passive MTU stiffness and neuromuscular coordination during locomotion (149). Because of the relatively slow time course for the contraction of the myofibroblasts inherent to fascia, this effect requires investigation over

a substantially long period of time, minutes to hours in length, and to date has not been investigated under these conditions.

Some indication of the potential discoveries awaiting further investigation of fascial contractility can be found in the investigations of sling systems throughout the body and the identification of differences in muscle activation patterns within these sling systems between runners with higher and lower RE (142,143). Sling systems are series of MTUs and their associated innervating fascia (142,143). These systems act to sequence muscle activation patterns within a kinetic chain and aid in cyclical movements. Perhaps unsurprisingly, the sling systems associated with improved RE, the deep longitudinal sling and lateral sling, are primarily comprised of muscles of the leg and lower back that control movement in the sagittal plane but also control movement in the frontal plane as well. The impact that increased fascial stiffness, and/or fascial contraction, within these sling systems could have on whole-leg stiffness, coordination of the co-contraction of biarticular muscles, passive stiffness, and therefore RE, could be significant and merits further investigation.

Multiple studies have demonstrated a significant negative correlation between the coactivation of biarticular muscles during the stance phase of running and cost of running (improved RE) (67,69,92,119,165). The mechanism responsible for this correlation has been identified as an increase in joint stiffness which minimizes hysteresis thus improving the utilization of elastic energy and increasing the RFD (32,67,81,90). Given the adaptive nature of fascia, and the recently acquired knowledge of fascial sling systems, it follows that repeated, specific, cyclical stress (running) to the connective tissue of the hip, knee, and ankle joint could stimulate fascial contraction aiding in the modulation of joint stiffness via increased passive stiffness. This same line of reasoning can be applied to the contribution of fascial contraction to



the coordination of the co-activation of biarticular muscles. Aiding in the effect of muscle co-activation, passive stiffness has also been tied to improvements in RE.

Increased passive stiffness of the MTUs of the shank has been negatively correlated with cost of running (improved RE) ( $r = -0.69$ ,  $p = 0.01$ ) while similar increase in vastus lateralis stiffness have been correlated with diminished RE (33,114). Interestingly, this contradiction between passive stiffness of the shank aiding RE while passive stiffness of the thigh musculature diminishing RE touches upon a very important, yet rarely discussed, aspect of lower extremity stiffness during running and its effect on elastic energy utilization. As mentioned previously, nearly all research directed at the interaction between stiffness and RE is focused upon stiffness of the shank (vertical, whole-leg, joint, and/or musculotendinous), specifically from the knee distally through the ankle during the stance phase of running. It has been demonstrated that increased stiffness during the stance phase of running is associated with improved RE which is indicative of improved utilization of elastic energy and thus lower hysteresis during running. Elastic energy returns from the MTUs of the shank have been estimated to comprise up to ~35% of total energy used during running. Accordingly, investigations into this portion of the lower extremity, which is almost exclusively only active during the stance phase, from initial ground contact through toe off, are important and justified. However, at ~60% of the gait cycle, the stance phase only represents a portion of the energy required to run. At higher velocities and with improved running mechanics the percentage of the gait cycle that GCT comprises shrinks. This leaves at least 40% of the gait cycle for the swing phase during which the lower extremity must be accelerated, from toe off at the termination of the stance phase, until the late swing phase when it is decelerated prior to ground contact. The swing phase represents a relatively small window of time during which a significant amount of work must be done to move a

significant mass (~ 14% of body mass). Increased stiffness of the elastic and contractile components of the lower extremity aid in the resistance of deformation during ground contact and in doing so limit potential energy loss, essentially improving the ‘spring’ in the leg-spring. This has been demonstrated repeatedly in the ankle and knee joints and the MTUs of the shank. But how do the elastic and contractile components of the leg function to utilize elastic energy during the swing phase? Though a relatively small amount of research has been dedicated to the question, and much more is required to properly evaluate the full potential, the answer may reside in the fascia of the leg and specifically a unique fascial structure, the iliotibial band.

### **3.7. Role of the Iliotibial Band in Running Economy**

Of all the distinct fascial structures of the human lower extremity, the iliotibial band (ITB) is perhaps the most prominent and presents the most promising subject for future investigations into elastic energy utilization in human locomotion. Surprisingly, the ITB is a poorly understood structure which has garnered most of its attention from research regarding ITB syndrome, anterior knee pain (patellofemoral pain), or 3D kinematics in runners (9,48–50).

The ITB presents as a thickened portion of the lateral fascia, the fascia lata, of the thigh segment of the lower extremity. Upon dissection the structure can be functionally divided into two separate sections based upon the muscles from which it inserts at its origin. The anterior ITB inserts entirely into the tensor fascia lata at its origin while the posterior ITB inserts into the gluteal muscles, primarily the gluteus maximus and gluteus medius (26). The ITB proceeds distally from its broad origins in the pelvis along the anterolateral aspect of the thigh to its insertions at the linea aspera, the upper border of the lateral epicondyle, the patella, the Gerdy’s tubercle, and the capsulo-osseous layer of the knee also referred to as the lateral femorotibial

ligament (26). As well as the anterior and posterior division, the ITB can be further dissected into three layers, the superficial, deep, and capsular-osseous (26). The deep layer is the site of insertion to the linea aspera, patella and Gerdy's tubercle (26). The insertion of the capsular-osseous layer consists of a pronounced ligament structure, labeled the lateral femorotibial ligament (26). Though the ITB can be carefully dissected to reveal a complex, multi-layered structure, much like the fascia found throughout the rest of the body its fibers appear to be continuous with the fascia, aponeurosis, and MTUs that lie in proximity (26). This is evident when examining the superficial and deep layers of the ITB as they appear to share fascia proximally into the pelvis, distally into the gastrocnemius and peroneals, anteriorly with the vastus lateralis and anterior aponeurosis, and posteriorly with the hamstrings.

Given its anterolateral orientation, origins in the pelvis and broad insertions distal of the knee, it is not surprising that the ITB has historically been classified as a knee stabilizer and received the limited attention that it has to this point. But, with the exception of recent research by Eng et al., this narrow view of the fascia of the thigh seems to be limited to humans. The elastic energy utilization capacity of the thigh fascia during running in multiple species of quadrupedal animals has been investigated and revealed great potential for energy saving from these structures (15). Early investigations of this concept estimated elastic energy recycled from the thigh fascia during quadrupedal running could constitute 17-25% of the energy used by the associated MTUs (15). More recent investigations, specifically those of Eng et al., have added weight to this theory via the first *in vivo* evidence of fascia energy storage, and most impressively, through comparative anatomy investigation and detailed modelling of chimpanzee and human ITB (36).

The first critical investigation by Eng and colleagues, en route to the broader exploration of the role of the ITB in RE, is their examination of the effects of biaxial strain on goat fascia lata (FL) (38). Similar to human ITB tissue, goat FL is composed of multiple layers with the collagen fiber from each layer predominantly oriented in a specific direction with respect to the body (38). These fibers run either longitudinally (superiorly and inferiorly parallel to the femur) or in a transverse orientation (anteriorly and posteriorly perpendicular to the femur) (38). The investigators hypothesized that perpendicular strain would preload the tissue and increase the stiffness, increase the strain energy stored in the tissue, increase the rate of loading in the stress-strain curve and that the two distinct layers/fiber orientations of the FL would behave differently under strain (38). Eng et al. demonstrated that perpendicular strain had no significant effect on the mechanical properties of either fiber orientation but did demonstrate significantly greater stiffness in the longitudinally oriented tissue (38). Though the investigators note limitations in these results, several important observations are made which prove important for subsequent investigation of human ITB function. Perhaps most importantly, Eng et al. demonstrated that the FL exhibited sufficient stiffness to transmit force in the longitudinal direction, aided by the relatively less-stiff transversely oriented tissue, while remaining compliant enough to allow stretch in the perpendicular direction (38). The investigators suggest that this characteristic of the FL *in vivo* would allow for the expansion of muscles as they contract while also being sufficient for elastic energy storage (38). Eng et al. also suggest that these findings demonstrate that the behavior of fascia is not strongly influenced by multi-axial strain and can therefore be modelled using conventional MTU models given the inclusion of fascia-specific data regarding bidirectional material properties (38).

Building upon Eng et al's., investigation of the effects of biaxial strain on fascia is their experimentation with the elastic energy storage capacity of goat fascia lata during locomotion *in vivo* (35). Currently, this type of invasive data collection is limited to animal subjects. Eng et al. used a combination of EMG and sonomicrometry to measure the muscular contraction patterns of the tensor fascia lata (TFL) and gluteal muscles and the stretch of the FL during walking and trotting on flat, incline, and decline in an attempt to quantify the potential elastic energy storage capacity of the FL *in vivo* (35). This particular instrumentation and combination of variables allowed the investigators to identify instances of peak muscular activity of the agonist muscles (TFL and gluteals) and peak stretch in the FL throughout the gait cycle. Importantly, this methodology also allowed for detailed characterization of stretch in specific regions and layers of the FL which provided important details regarding the stretch and recoil of the FL throughout the swing phase, specifically. The study divided the goat FL into two sections. The so named proximal region of the FL was the portion which inserted into the TFL while the distal portion inserted into the gluteal muscles (35). The researchers demonstrated that the distal goat FL exhibited characteristics during walking and trotting that strongly suggest a potential to store elastic energy during the stance phase which would then be utilized during the swing phase (35). The data clearly exhibits a consistent pattern of longitudinal strain and stretch in the distal FL during the stance phase, in conjunction with contraction of the gluteal muscles, and shortening in the swing phase (35). The study also produced data that suggests a potential for the proximal FL to store elastic energy in the early swing phase which would subsequently be released in the late swing phase and initial stance phase (35). The results indicate that the proximal FL was passively stretched during stance phase via hip extension and then, potentially, further stretch/strain occurred coincident with the isometric contraction of the TFL during the early swing phase (35).

Data regarding the proximal FL was plagued with inconsistency between subjects, and few total subjects, and was therefore inconclusive (35). Though the arrangement of the musculature that inserts into the goat FL differs from human ITB, the *in vivo* experimentation provided valuable insight into muscle activation and strain patterns that can be applied to aid in the modelling of the human ITB.

Eng et al. continued in their investigation of the elastic energy storage capacity of the ITB with the aid of a complex model of the human lower extremity. This investigation was flanked by a companion study to test the origin of the ITB in order to derive a better understanding of its function as an energy saving adaptation. Eng et al. took multiple important steps to integrate the results of their previous experimentation, cadaveric specimens and a recalculation of the force-generating capacity of the TFL and gluteus maximus (GM) in order to overcome the limitations of previous models. The researchers based their redesigned model of the force-generating capacity of TFL and GM on dissected samples from three human cadavers (37). From these specimens total mass, fascicle length, surface pennation angle and cross-sectional area were measured. In order to accurately model the three dimensional anatomy and moment arms of the TFL and ITB (TFL-ITB) MTU and the GM and ITB (GM-ITB) MTU, five additional cadavers were examined (37). From this set of specimens, moment arms for flexion and extension, internal and external rotation, and abduction and adduction of the hip, and knee flexion and extension were measured via the tendon excursion method (37). Additionally, a motion tracking system was employed to capture joint angles concurrently with MTU length change in order to determine the joint angles at which the ITB exhibited stretch (37). Based upon musculoskeletal modelling by Arnold et al. (6), and adjusted based upon their earlier experiments with goat FL, Eng et al. chose to model the anterior and posterior ITB as individual MTUs acting upon five

separate lines of pull, two of which represent the TFL-ITB MTU and the remaining three representing the GM-ITB MTU (37). Each of the five major lines of pull included in the model were consistent with pathways of ITB force transmission included in previous models and verified by the investigators measurements of moment arms and MTU excursion (37). Because previous models of MTUs incorporated the force-length curves of tendons and there is a significant difference in the behavior of tendons and fascia, the investigators generated new force-length curves based upon the stress-strain data reported by Derwin et al. and included tendon slack length as an additional parameter in order to improve the accuracy (30). To calculate elastic energy storage of the ITB, periods of peak MTU length and muscle activity were identified throughout the running and walking gait at speeds of 2, 3, 4, and 5 m/s. These instances during the running and walking gait were assumed to be moments of peak strain (37). The model was then adjusted to separate the calculated MTU lengths into ITB lengths and muscle lengths. Finally, following computer simulations utilizing the new model to generate ITB strain estimates, elastic energy (E) was calculated using the formula:

$$E = \frac{1}{2}F_{ITB}\Delta L_{ITB}$$

The results revealed two distinct patterns of maximal stretch and EMG activity between the TFL-ITB and GM-ITB (37). The TFL-ITB exhibited peak stretch during the late stance phase during walking and early swing phase during running, during which the appendage is extended at the hip, flexed at the knee and coincident with EMG activity of the TFL (37). The GM-ITB showed peak stretch during the late swing phase during which the hip is flexed and the knee extended and coincident with EMG activity of the GM (37). Eng et al. report significant stretch of the anterior and posterior ITB with peak strains during sprinting as much as 5.6% and 12%

respectively (37). According to the investigators this represents significant stretch of the ITB caused by the coincident of passive stretch from hip and knee flexion and extension, and the contraction of TFL and GM. Estimates of energy storage range from 2.35 J in the anterior ITB to 12.1 J in the posterior ITB during sprinting (37). Interestingly, the investigators experimented with augmentations of the stiffness of the ITB to determine the effects of a stiffer ITB on elastic energy storage potential. The experimentation demonstrated a significant decrease in the energy storage of the anterior and posterior ITB throughout all speeds, with peak losses estimated at 57% and 66% for the anterior and posterior ITB respectively during sprinting (37). The investigators identify several limitations in their study but maintain their models relative accuracy and the results strongly indicate that the ITB stores elastic energy at all walking and running speeds tested (37). Chief amongst the limitations addressed by the investigators is the use of cadavers to create force-length curve models for the ITB. Eng et al. acknowledge the potential for significant difference exists between the cadaveric specimens used and the ITB of a healthy human but suggest that the main difference, stiffness, presents a negligible difference in resultant energy storage capacity. The investigators also address the potential for the modeled force-velocity curves of the TFL and GM to be inaccurate as they were completely ignored in the simulation. It is suggested, however, that the TFL and GM may contract nearly isometrically during instances of maximal force generation while running, similar to the gastrocnemius, as has been previously discussed (37). If this is the case, force-velocity characteristics become a moot point and the accuracy of the model is maintained. Also, the investigators acknowledge that while the study demonstrated the storage of elastic energy in the ITB it did not properly test whether or not this stored energy effectively reduces the energetic cost of walking or running (37).



Eng et al. also conducted a comparative anatomy study in order to test the theory that the ITB is a specialized structure that evolved as an adaptation to improve bipedal economy. The study utilized the same methods as the previously mentioned ITB study with the addition of four chimpanzee cadavers allowing for the comparison of the bipedal human structure to the primarily quadrupedal chimp structure (36). The results strongly suggest that, when compared to chimpanzee FL, the human ITB is a structure specialized for elastic energy storage (36). The investigators' assertions are buoyed in the fact that the human GM is significantly larger than the chimpanzee GM and therefore transmits significantly more force. Given the previously discussed capability of the ITB to store elastic energy, this development of GM size suggests the fascial structure may have developed as a means by which to exploit this greater force-generating capacity and improve economy (36).

Collectively, the investigations by Eng et al. strongly suggest multiple new concepts regarding the human ITB that should stand as beacons for future research exploring fascia and exercise economy. This series of studies established that the ITB's mechanical properties are not strongly influenced by multi- or bi-axial strain and instead function in such a way as to transmit force primarily along a single axis while also retaining the ability to transmit lateral force and deform along other axes simultaneously (38). The *in vivo* study clearly demonstrates that the FL of a goat, which shares functional characteristics with a human ITB, can store significant amounts of elastic energy during locomotion (35). The human cadaver and musculo-skeletal modeling study integrated the findings of the preceding investigations and established a clear pattern of muscle activity and fascial mechanics across a range of velocities (37). These findings in turn strongly suggest the ITB is capable of storing a significant amount of elastic energy which logically would be returned during the gait cycle aiding in walking and running economy.

(37). This assertion is supported with the comparative anatomy study which establishes strong anatomical evidence suggesting the ITB is a structure specialized for exercise economy (36).

These findings, while focused solely on the ITB, have much broader implications and, at the very least, support the assertion that fascia plays a crucial role in locomotion and the coordination of movement.

#### **4. Discussion**

Running economy is a crucial component of running performance and is determined by a multitude of physiological, biomechanical, and neuromuscular variables (42,46). The capability of a runner to successfully utilize elastic energy in order to decrease the metabolic cost of running is one of the most important components of RE (27). Stiffness on all levels, whether it is the global vertical stiffness of the runner, whole-leg stiffness quantifying the compliance of the lower limb or MTU and aponeurosis stiffness, has been demonstrated to be a critical component of elastic energy utilization. Specifically, the majority of research reveals the importance of stiffness lies in the reduction of the energetic cost during the weight-bearing stance phase of running. Of greatest importance during the stance phase of running is the stiffness of the ankle, MTUs of the shank, and knee.

Currently, the measurement of stiffness, in the context of running, is most commonly performed via computer-generated musculoskeletal models or multi-camera motion capture systems that utilize inverse dynamics. Inverse dynamics, while currently being the most accurate method to quantify stiffness, has inherent flaws and fails to produce results that accurately represent the biomechanics of running. Improvements upon musculoskeletal modelling, and the inverse dynamics calculations inherent to them, have increased the accuracy of this methodology

but to date have failed to produce a robust method capable of accurately describing the kinematics and kinetics of the lower limb during running. Of the multiple assumptions inherent to inverse dynamics, the four assumptions identified in this study: 1) joints are frictionless pin-joints 2) the segments of the leg are rigid 3) the mass of each segment is concentrated at the center, and 4) there is no co-contraction of agonist and antagonist muscles, refer directly to properties of the lower extremity that have been demonstrated to be directly associated with vertical stiffness, whole-leg stiffness, joint stiffness, or an anthropometric determinant of RE. In order for future musculoskeletal modeling techniques to improve in accuracy and remain as a valid tool for the investigation of the more subtle aspects of running biomechanics, substantial effort is required to negate these counterproductive assumptions. This proposal has been suggested previous in the pertinent literature but to date no viable, affordable solution has been created.

Important questions regarding the biomechanical properties of the tissues of the lower limb remain unanswered because of the limitations of the non-invasive techniques used to study them. This void of knowledge limits speculation about the underlying mechanisms of RE to data derived from investigations whose results are limited in their validity due to the established shortcomings of the accuracy of their methodology. While research has clearly demonstrated that increased vertical stiffness and whole-leg stiffness are correlated to improved RE, limitations in the investigation of MTU and joint stiffness due to limitations in methodology, leave many questions unanswered. The quantification of vertical stiffness and whole-leg stiffness function to establish the relationship between stiffness and RE but do little to elicit the underlying mechanisms of the relationship between the two. No clear answers regarding the optimal levels of stiffness at the vertical, whole-leg, joint, or MTU level are known. Nor is there a clear

consensus on the acute or long-term effect of training interventions, across all modalities, on any metric of stiffness or the subsequent effect these potential alterations have on RE.

The recent developments in the research of the contractility of fascia and ability of fascia to modulate stiffness in the lower extremities of humans presents new possibilities in the investigation of the underlying mechanisms governing musculotendinous and joint stiffness (149). It is suggested that fascia may play a significant role in the modulation of passive stiffness of the MTUs of the lower limb. A growing body of research exploring fascial sling systems and their role in exercise economy supports the concept that fascia may play a vital role in the coordination of the co-contraction of biarticular muscles during the stance phase of the gait cycle, which has been associated with increased stiffness and improved RE. Specific exploration of the ITB has demonstrated its potential as a significant contributing factor in the utilization of elastic energy during the running gait cycle. These investigations have also demonstrated the importance of differentiating the force transmission and elastic energy storage characteristics of fascia from that of ligament, tendon, and muscle tissue. Not only should fascia continue to be explored in greater depth in order to derive its functional role in exercise economy but also in order to better characterize its biomechanical behavior. A greater understanding of the biomechanical behavior of fascia will significantly improve future musculoskeletal modelling efforts which in turn will improve the accuracy of future investigations of elastic energy and passive force generation. The investigations of fascia and the ITB have begun to shed light upon long-ignored aspects of RE but also revealed the depths of the lack of knowledge presently of the role of the hip joint in RE, the role of fascia in the coordination of co-contraction (throughout the body), and the trainability of fascia and fascial sling systems.

Given the revelations of more current research exploring fascia and improved investigative methodology, supported by a vast body of prior inquiry into the biomechanics of running, it is apparent that future research should be directed at investigating these topics. The most important amongst these research results is the light shed upon the role fascia plays in elastic energy utilization. The identification of distinct differences in the activation patterns within the sling systems of individuals with high versus low RE, coupled with the demonstration of the elastic energy storage capacity of fascial structures, speaks to the potential for further research. These revelations are further buoyed by the demonstrated ability of fascia to adapt its biomechanical characteristics via contraction and aid in the coordination of movement due to its complex innervation. The future of running economy research should be directed at further exploration of the fascial structures and sling systems throughout the body.

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## **Review of Pertinent Literature**

### *Introduction*

It is well established that running performance is primarily determined by maximal oxygen consumption ( $VO_{2max}$ ), fractional utilization of  $VO_{2max}$  ( $\% VO_{2max}$ ), anaerobic threshold (LT) and running economy (RE) (53,54,102). Running economy relies on multiple physiological components, such as the cardiovascular and neuromuscular systems, that drive biomechanical factors: force, power, ground reaction forces, stride frequency and length, and ground contact time (32,89,92,93). The combination of these factors collectively determine the metabolic efficiency of an individual while running (32,89,92,93). A growing body of literature has illustrated the effects of a multitude of interventions targeted at improving RE in both trained and untrained populations (55,67,75,77,78,89,92,93,115,118). Of these experiments, the vast majority that focus on biomechanical variables identify stiffness, either musculotendinous, whole-leg or vertical, as a major contributing factor to RE. While this research has begun to explore the influence of stiffness upon the economy of exercise, the scope of knowledge on the subject has yet to fully explain many of the underlying mechanisms. This review of literature explores running economy and stiffness to establish a base of knowledge in order to clarify future investigation into aspects of the subject that lack rigorous investigation including: the modulation of stiffness, the role of fascia, and the role of the thigh and hip in the utilization of elastic energy.

### *$VO_{2max}$ and Fractional Utilization of $VO_{2max}$*

Maximal oxygen uptake ( $VO_{2max}$ ) is defined as the maximum rate oxygen can be consumed by the body and utilized for exercise (7,8,44). Maximal oxygen uptake is often described as the upper limiting factor for aerobic performance and is considered the main

determinant of performance in short running events from 1000 meters to 3000 meters in length as well a contributing factor to performance in longer events. A consensus of the literature concludes that  $\text{VO}_2\text{max}$  is limited by the performance of the pulmonary system, maximum cardiac output, and oxygen carrying capacity (7). Maximal oxygen uptake ( $\text{VO}_2\text{max}$ ) is most often measured via graded exercise test (GXT) until voluntary failure. Multiple validated variants of GXT are utilized in the literature. Most variations measure expired gases during a running protocol that is concluded once  $\text{VO}_2$  consumption ceases to increase with increased workload or the subject terminates the test. Maximal oxygen uptake ( $\text{VO}_2\text{max}$ ) can also be estimated from submaximal testing in which  $\text{VO}_2$  values are used to calculate  $\text{VO}_2\text{max}$ .

The fractional utilization of  $\text{VO}_2\text{max}$  ( $\% \text{VO}_2\text{max}$ ), or the sustainable percent of a subject's  $\text{VO}_2\text{max}$  for a given distance is a key determinant of endurance performance. While  $\text{VO}_2\text{max}$  can be considered to establish a maximum ceiling for performance potential,  $\% \text{VO}_2\text{max}$ , along with RE, establishes how much of that potential is realized. Lactate threshold and  $\% \text{VO}_2\text{max}$  are often closely related. Lactate threshold (LT) is defined as the point during exercise at which blood lactate accumulates at a rate exceeding the rate at which it can be removed (7). LT is measured via direct blood sampling during GXT. The LT is determined following lactate analysis and defined as the velocity at the lactate inflection point when blood concentration rapidly increases. Maximal oxygen uptake ( $\text{VO}_2\text{max}$ ) and  $\% \text{VO}_2\text{max}$ , in conjunction with LT, define an upper limit of sustainable, submaximal performance but fail to complete the picture of how long a given submaximal (below LT) effort can be maintained. The remaining element, running economy (RE), defines the metabolic cost of submaximal performance.

## *Running Economy*

Running economy, defined as the steady-state oxygen consumption ( $\text{VO}_2$ ) at a given speed, has been widely identified as a key component in running performance (108) specifically in mid-distance and long-distance running (7,76). Multiple methods exist for the measurement of RE and despite variations in methods, protocol, and analysis, a consensus culminating in representative values has been established allowing for a degree of accuracy in the comparison of individuals given established parameters (32). Running economy is the complex product of the interaction between neuromuscular, biomechanical, cardiorespiratory, and metabolic components unique to the individual (6,13,21). Research has shown the efficacy of a variety of interventions to positively affect these components and improve RE (6). While this review is primarily focused on the biomechanical and neuromuscular components of RE, it is important to clarify all aspects of this valuable component of running performance in order to better understand the crucial contribution of musculotendinous stiffness to RE.

Multiple methods for evaluating RE at submaximal running speeds have been employed throughout the literature but the majority use the collection of expired gases in order to measure the volume of oxygen consumed ( $\text{L}\cdot\text{min}^{-1}$  or  $\text{mL}\cdot\text{kg}^{-1}\cdot\text{min}^{-1}$ ), expressed as a percentage of  $\text{VO}_2\text{max}$  during a given bout of submaximal running (14,28). Various submaximal speeds have been utilized in the measurement of RE with the most common being 268 m/min ( $4.47\text{ m}\cdot\text{s}^{-1}$ ) or six-minute mile pace (32). Additionally, RE has also been measured as the  $\text{VO}_2$  consumed while running a given distance and then normalized for body weight ( $\text{mL}\cdot\text{kg}^{-1}\cdot\text{km}^{-1}$ ) (21,32,51,68). Research has also suggested that alternative means of measuring RE may be more accurate and more representative of the multidimensional nature of RE. Fletcher et al. (28) compared the standard method of measuring RE via expired gases with that of the same methodology plus the

measurement of gross caloric unit cost ( $\text{kcal}\cdot\text{kg}^{-1}\cdot\text{km}^{-1}$ ) assessed via analysis of respiratory exchange ratio (RER). The researchers calculated this caloric unit cost as the product of the volume of oxygen consumed in liters per minute ( $\text{VO}_2$ ), the caloric equivalent ( $\text{kcal/L}$ ), velocity, body mass, and distance, in this case 1 kilometer. This alternative method allows for the assessment of the substrate being utilized at a given steady state velocity and is suggested to be a more accurate measurement of actual energy cost as the energy derived from a given quantity of oxygen is dependent upon the substrate being metabolized (28–30,100,117). Running economy has also been assessed in trained distance runners using a multi-speed submaximal treadmill test in which subjects performed two six-minute efforts separated by a five-minute rest period (128). The first of these six-minute trials was performed at  $16.1\text{ km}\cdot\text{h}^{-1}$  (six-minute mile pace) and the second trial at the subject's current 10-km race pace (128). This approach provides the added benefit of a more detailed analysis of substrate utilization via the comparison of the steady-state RER values between trials thus resulting in a more accurate assessment of RE as well as  $\% \text{VO}_2\text{max}$  (128).

#### *Determinants of Running Economy*

Investigations into the determining factors of RE have illuminated several physiological components that may have significant impacts on RE and therefore running performance. These factors include blood lactate concentration (LT), substrate metabolism, body core temperature, heart rate, and minute ventilation (4,62,85,93,108). The body of literature generally agrees that body core temperature, minute ventilation, and heart rate have some measurable effect on exercise economy. However, the change in  $\text{VO}_2$  demand due to variation of these measures is too small to be considered a major contributing factor (4,86,96,102,124). A greater capacity for lactate buffering theoretically allows for a higher velocity at LT, thus improving  $\% \text{VO}_2\text{max}$  at

LT. This, coupled with an increased functionality of skeletal muscle mitochondria, increased fat metabolism, and a lower dependence on muscle glycogen account for the most significant physiological contributions to improved RE (6,28).

Several biomechanical factors have been identified as contributing to superior RE. These include rate of force development (RFD), stride frequency, ground contact time (GCT), swing time, and stride length (25–27,31,35). Multiple anthropometric variables have been investigated in association with RE, however, only mass distribution and Achilles tendon moment arm length have been shown to benefit RE (6,32,50,64,68,114,129). Multiple studies examining the biomechanical properties of East African distance runners correlated a tendency of mass distributed proximally in the lower extremities to increased RE (32,64,68). These findings have been supported by multiple studies examining the effects of distal loading of the lower extremities (71,88,98). The work of Scholz et al. (113), Hunter et al. (50), and Barnes et al. (6) showed very strong correlations between a short Achilles tendon moment arm, which would effectively increase musculotendinous stiffness and therefore the utilization of elastic energy by limiting hysteresis with improved RE. This specific aspect of anthropometry is also directly related to joint stiffness and will be discussed further. A great deal of research has also been directed at investigating the intersection between gait characteristics and RE. A consensus of the findings of this work strongly suggests that RE is closely associated with self-selected stride length and frequency (4,15,57,130) that generates the least amount of vertical oscillation (15,40,42,122,130) and therefore generates less vertical GRF and more horizontal GRF (18,42,89,91).

## *Stiffness*

Along with investigations of the physiological and biomechanical determinants of RE, a considerable amount of research addresses the neuromuscular components of RE. Specifically, stiffness has received special attention in these investigations as it has been identified as the potential mechanism driving several other variables, both neuromuscular and biomechanical, namely RFD, GCT, elastic energy utilization, and passive forces. These variables have been linked to superior RE (32,68,89,93,106,107) with the relationship between stiffness and elastic energy utilization strongly suggested as the most significant determinant of RE.

Stiffness can be defined as a resistance to deformation when force is applied to an object (20). This measurement of deformation can be applied to the lower extremities of humans on multiple levels including vertical, whole-leg, joint, and musculotendinous stiffness (1,6,20). Stiffness, in reference to biomechanics, is based upon Hooke's Law and is most often modelled as a spring. Hooke's Law serves as a simple approximation of the action of an elastic body when force is applied to that body and states that: the force ( $F$ ) required to deform the elastic body is equal to the stiffness ( $k$ ) of the elastic body multiplied by the amount of deformation ( $x$ ) when  $F$  is applied. The equation  $F=kx$  can be used to calculate stiffness when the force applied to that object and the resultant deformation is measured.

Utilizing Hooke's Law, multiple models have been developed in order to study biomechanical variables of the lower extremities. The most common of these models is the spring-mass model which describes a mass which is representative of the center of mass resting above a weightless spring that is representative of the lower extremity (20). Forces in the form of GRF compress the spring, analogous to the eccentric initial phase of ground contact. Mechanical energy is stored within the spring, and then released as the spring recoils which is analogous to



the combination of concentric muscle action and elastic energy recoil during the propulsive phase of ground contact (20,73,87). The degree to which the lower extremity resists compression (deformation) is dependent upon the stiffness of that element, specifically whole-leg stiffness which in turn is dependent upon muscle, tendon, and aponeurosis stiffness (1,3,12,82). Consequently, the amount of elastic energy return, or utilization, during the propulsive moments of the stance phase of running is positively correlated to stiffness and ultimately RE (2,3,20).

Inverse dynamics, a technique which allows for the calculation of multibody dynamics, has been used extensively to analyze the kinematics and kinetics of running and is the basis upon which modern computer-generated modelling is built (27,62,101,116,125). Inverse dynamics utilizes the Newton-Euler equations for force (*force = mass x linear acceleration;  $F = ma$* ) and moment (*moment = mass moment of inertia x angular acceleration;  $M = I\alpha$* ) and a theoretical link-segment model of the lower extremity to calculate the joint moments at the ankle, knee and hip to derive the kinetics of the leg (9). Using this technique joint stiffness can be calculated along with the work and power of each segment and joint and the effect of velocity upon these parameters (9).

While inverse dynamics is a valuable tool that is widely utilized, it is based upon several assumptions that, at best, allow for useable approximations and is sensitive to the accuracy of the instrumentation and skill of the technicians conducting the data collection (46,47,72). Of the assumptions inherent to inverse dynamics, several present direct conflicts with the accuracy of this technique to describe the kinetics of the leg during running. These include: 1) the joints are frictionless pin-joints 2) the segments of the leg are rigid 3) the mass of each segment is concentrated at the center, and 4) there is no co-contraction of agonist and antagonist muscles. A basic understanding of anatomy coupled with a wealth of research clearly illustrates that these

assumptions are obviously not representative of reality and challenge the accuracy of an inverse dynamics analysis of the human leg during running (11,27,36,62). Several studies have attempted to improve the accuracy of inverse dynamics via optimization algorithms (27), smoothing algorithms (36) and least-squares estimation computation (11,62). These attempts at improving the precision of inverse dynamics analysis provide valuable insight into the nature of the error present in the technique. Classic inverse dynamics calculation starts at the most distal segment (foot contact with a force plate) and proceeds proximally to calculate all successive parameters. Because of this order of operations, and the assumptions necessary for the calculation, constant error begins to compound and creates substantial error in the calculation of all variables and the accumulation of residual forces and torques (11,62). Due to this phenomenon, joint reaction forces, joint stiffness, and work performed at the joint are often inaccurately estimated.

### *Vertical Stiffness*

Vertical stiffness is a calculation of the vertical excursion of the center of mass. The calculation of vertical stiffness has evolved through multiple methods as instrumentation has become more advanced and made more in-depth measurements of kinetics possible. The initial method to calculate vertical stiffness was formulated by McMahon et al. (74)(107) and stated that vertical stiffness was equal to the product of mass and the square of the natural frequency of oscillation ( $k_{vert}=m\omega^2$ ). In this formulation, McMahon derived the natural frequency of oscillation ( $\omega$ ) from contact time and vertical velocity which was calculated from a vertical force and contact time curve created by data gathered from the utilization of a force plate (74)(107). Cavagna et al. (16)(22) expounded upon McMahon's initial method, also using a force plate to calculate a natural frequency of oscillation ( $\omega$ ), but instead making the calculation from ground

(force plate) contact time. McMahon and Cheng (73)(106), in order to create a simplified model for human running and the relationship between velocity and stiffness, calculated stiffness from maximum vertical force and maximum vertical displacement. Specifically, the McMahon and Cheng method states that vertical stiffness is equal to the change in vertical force divided by the change in vertical displacement ( $\Delta f_y/\Delta y$ ) at the instance of maximum vertical force and maximum vertical displacement during the mid-stance phase of running (73)(106). Morin et al. (87)(125) attempted to simplify the popular model of McMahon and Cheng by calculating vertical stiffness from mass, ground contact time and aerial time, the latter two variables being measured via pressure sensors worn by the experimental participants in their shoes. The researchers used this data to create a sine wave-based force-time curve model from which vertical stiffness was calculated (87)(125). They concluded that their proposed method resembled contemporary force plate derived stiffness data and offered a simple method that could be employed beyond the confines of a laboratory.

### *Whole-leg Stiffness*

Whole-leg stiffness, also frequently referred to as leg stiffness or leg spring stiffness, is a measurement of the stiffness of the whole leg when the three-segment limb is considered as a single spring-like structure. McMahon and Cheng's (73)(106) work on establishing a simplified model for human running in order to better understand the relationship between velocity and stiffness, which resulted in a popular method for calculating vertical stiffness, also resulted in an oft-used method for calculating whole-leg stiffness. The method the researchers developed employed the vertical force data from a force plate while the change in whole-leg length was derived from velocity, instantaneous leg segment length, whole-leg angle at the moment of initial impact and vertical excursion of the center of mass (73)(106). McMahon and Cheng (73)(106)

then expressed leg stiffness as a ratio of maximum vertical force to maximum change in leg length. Morin et al. (87)(125) also applied their previously mentioned simplified method for modeling running to the measurement of whole-leg stiffness. Foregoing the force plate, Morin et al. (87)(125) used running velocity, measured with a radar gun, and initial leg length to calculate whole-leg stiffness from the maximum change in leg length. Arampatzis et al. (2)(3) conducted a study compare the mathematically calculated leg length changes with those measured kinematically. Arampatzis and colleagues employed a force plate (sampling at 1000 Hz) and two high-speed cameras (sampling at 120 Hz) and a system of reflective markers to conduct data collection. Following this, whole-leg and vertical stiffness were calculated using formulas already established previously in research (73,74)(106,107): leg stiffness ( $K_{leg}$ ) is equal to maximum vertical ground reaction force ( $F_{z\ max}$ ) divided by leg length change ( $\Delta L$ ); vertical stiffness ( $K_{vert}$ ) is equal to  $F_{z\ max}$  divided by vertical displacement of the subject's center of mass ( $\Delta y$ ). The leg stiffness values reported by Arampatzis and colleagues (2)(3) far exceeded those calculated by any previous study employing the established mathematical stiffness calculation methods. Ultimately the researchers argue that this difference exists due to a tendency for overestimation of the excursion of the center of mass that is inherent when using the method made popular by McMahon and Cheng (73)(106).

### *Joint Stiffness*

Joint stiffness, of a single joint, is the ratio of joint moment to the angular displacement of that joint. The most common method employed for the calculation of joint stiffness uses the same data collection methods as contemporary studies of whole-leg and vertical stiffness (ground reaction force data from force plate(s) and digitized kinematic data from high-speed video footage in conjunction with a marker system indicating anatomical landmarks) but also utilizes

inverse dynamics for the calculation of joint moments (38,52,61,121)(61,81,90,162). From this data joint stiffness is then calculated as the change in joint moment ( $\Delta M$ ) of a given joint divided by the change in joint angle ( $\Delta\theta$ ) of that joint (38,61,121)(61,90,162).

Arampatzis et al (2)(3) attempted to calculate joint stiffness along with leg stiffness in order to accurately evaluate the relationship between velocity and stiffness in human running. The researchers used the same kinematic data gathered for the other elements of the study and modeled the ankle joint and knee joint as “rotational springs”. The calculation was expressed as a ratio of negative work to change in joint angle: ankle stiffness ( $K_{\text{ankle}}$ ) is equal to two times the negative mechanical work of the ankle joint ( $W_{\text{ankle}}$ ) divided by the change in ankle angle ( $\Delta\theta_A$ ); knee stiffness ( $K_{\text{knee}}$ ) is equal to two times the negative mechanical work of the knee joint ( $W_{\text{knee}}$ ) divided by the change in knee angle ( $\Delta\theta_K$ ). This method later received criticism on multiple levels, which claimed it was irreproducible and has not been utilized since (38)(61). All other studies employ standard inverse dynamic calculation methods or modelling software which calculates joint stiffness, among other variables, using inverse dynamics (52,61)(81,90).

### *Muscle-tendon Unit Stiffness*

Musculotendinous (MTU) stiffness is a key contributor to whole-leg stiffness and is positively correlated to RE (1,3,12,73,80,82)(2,4,19,106,118,120). The most common technique for the measurement of MTU stiffness is the free oscillation technique (22,126,127)(31,168,169), though alternative methods exist for the measurement of tendon stiffness (65)(94) and new methods continue to be explored and validated (17,58,94)(23,88,133). The free oscillation technique models the MTU system, consisting of the contractile element (CE), series elastic element (SE) and parallel elastic element (PE), as a damped mass-spring system (22)(31). The

technique involves loading a MTU in a given body segment and then perturbing that segment which results in damped oscillations which are measured and used to calculate stiffness (22)(31).

Tendon stiffness has been measured in isolation to MTU stiffness and has also been positively correlated to RE (29)(43). Tendon stiffness has most commonly been assessed via ultrasonography (56,59,60)(85,86,89) though more advanced techniques have also shown promising degrees of validity (18,65)(24,94). The use of ultrasonography to measure tendon stiffness employs the standard stiffness formula ( $F=kx$ ) by deriving force from joint torque and displacement from tendon excursion assessed using ultrasonography (29,56,59,60)(43,85,86,89). Similar techniques have been applied to evaluate the passive stiffness of both tendons, muscles, and MTUs. Passive stiffness has been evaluated using the free oscillation technique (24)(33), an isokinetic dynamometer (34,104)(51,141), shear wave elastography (ultra-fast ultrasound) (65,79)(94,114) and myotonometry (17,65)(23,94). The measurement of passive stiffness using the free oscillation technique is identical to the method used for measuring MTU stiffness as described above. To elaborate, the subject being evaluated has the MTU in question loaded with successively heavier weights within a range (ex. 5-40% (24)(33)) of their established maximal isometric contraction of that MTU. Each of these loads are then perturbed and the resulting oscillation of the MTU is measured using a force plate. Stiffness is then calculated using the measured oscillation pattern and the equation ( $m \frac{d^2x}{dt^2} + b \frac{dx}{dt} + kx = F_0$ ) which describes the motion of a dampened spring (24)(33). The evaluation of passive stiffness via isokinetic dynamometer employs software proprietary to the manufacturer of the dynamometer that calculates stiffness from force and angle data collected during each experimental trial (104)(141). Shear wave elastography measures passive stiffness by generating an acoustic radiation force in the form of a focused pulse from a hand-held unit and then measures the speed

of the propagation of the shear wave as it reacts to the stiffness of the tissue upon which the beam was focused. The shear wave speed is then interpreted as a representative of tissue tensile strength because of its correlation to elastic moduli (65,79)(94,114). Myotonometry measures passive stiffness in a similar manner to the free oscillation technique in that the calculation of stiffness is derived from a measurement of damped oscillation but differs in its method of causing the oscillation. The most recent development in myotonometer, the MyotonPRO (Myoton AS, Tallinn, Estonia) is a hand-held digital palpation device that generates a mechanical force into a targeted tissue and measures the resultant damped oscillations with a built-in accelerometer and calculates stiffness as the ratio of resistance to external force and the maximum displacement of the target tissue (17,65)(23,94).

#### *Relationship Between Stiffness and Running Economy*

Because of their dominance in distance running at an international level, East African runners have been studied in detail in order to evaluate physiological or biomechanical mechanisms that would explain their success (64,84,106,107). For this reason, East Africans have also become a model for the evaluation of the contribution of RE to running performance (64). Though many questions still remain regarding the inherent mechanisms of East African dominance, one observation, short GCT, which is correlated to superior RE, has led to further investigation of GCT and stiffness of muscle-tendon unit (MTU) and aponeurosis (10,45,83,84,105). Subsequent investigation has demonstrated that increased stiffness of MTU and aponeurosis has been negatively correlated with the energetic cost of running (20,77,92,93,118).

Greater musculotendinous stiffness is positively correlated with improved RE via improved elastic energy utilization and decreased GCT (20). In a study conducted by

Paavolainen et. al. (93), an increase in leg stiffness was interpreted as a key mechanism in the observed improvement in RE due to the significant decrease in GCT in conjunction with the significant increase in peak sprint velocity, and the concurrent improvement in jump testing. This connection between GCT and stiffness is indicative of an improvement in the efficiency of the stretch-shortening contraction (SSC). Stretch-shortening contractions occur when muscle-tendon units are actively lengthened and then concentrically contracted as in running or a counter movement jump (20). As the muscle-tendon unit is stretched, elastic strain energy is stored in the connective tissue and, upon concentric contraction, can add to the force and power generated by the contraction (33). The degree to which this elastic energy is utilized or dissipated during the SSC is dictated by multiple physiological and biomechanical properties and is analogous to the phenomena of hysteresis (33). It follows that a reduction in the hysteresis, or parasitic loss of energy, in a SSC, specifically running, would greatly benefit the utilization of elastic energy and would therefore improve distance running performance via an improvement in RE (20).

The connection between stiffness and RE has been demonstrated repeatedly in training studies in which resistance training interventions have been applied to distance runners, across a broad spectrum of experience, in order to assess the effect of the intervention upon the established determinants of running performance

(33,37,39,55,66,67,75,77,78,92,93,99,115,118). Piacentini et. al. (99) reported significant increases (13%,  $p < 0.05$ ) in stiffness following a 6-week training protocol. The study employed two different concurrent training (CT) modalities: maximal strength training (HRT) (performed at 85-90% of estimated 1RM), and resistance training (RT) (performed at 70% 1RM) (99). The reported increase in stiffness was observed only in the (RT) group while the (HRT) group and comparison group showed no significant effect on stiffness (99). Following a 14-week training



intervention consisting of isometric ankle plantar flexion, Albracht et al. (1) showed a significant increase in tendon-aponeurosis stiffness of the triceps surae (15.8%,  $p < 0.01$ ) which along with an increase in RE (4.35%,  $p < 0.05$ ) resulted in significant group-by-time interactions with  $\text{VO}_2$  consumption ( $F = 5.2$ ,  $p = 0.03$ ) and energy cost ( $F = 5.4$ ,  $p = 0.02$ ). The experimental group showed a 4.2% ( $p < 0.05$ ) reduction in energy cost per body weight and a 4.8% ( $p < 0.05$ ) reduction in  $\text{VO}_2$  consumption while the control group showed no change.

Mirroring the findings of Piacentini et al. (99), a study of the effects of concurrent plyometric (PLY) training conducted by Spurrs and colleagues (118) reported significant increases in stiffness for the experimental group following a 6-week intervention. Results were reported for each leg individually and showed a 14.9% ( $p < 0.05$ ) and 10.9% ( $p < 0.05$ ) increase in stiffness for the left and right legs, respectively. Though measurements of MTU stiffness in response to concurrent training (CT) requires further research, the literature strongly suggests that CT incorporating plyometrics (PLY) significantly increases MTU stiffness and that the mechanism for this enhancement is most likely the high-velocity eccentric loading occurring during the SSC characteristic of PLY training (33,118). An incredible wealth of literature examining the effect of concurrent training interventions incorporating a multitude of modalities has been generated in the last four decades. The consensus of this repository concludes that, while some interventions are more effective than others, interventions that improve the neuromuscular characteristics of the lower limb in regards to pre-activation of the agonist and antagonist MTUs, and MTU and joint stiffness have the greatest beneficial effect on RE (5,13).

Of the variables examined in the investigation of RE, GTC receives a lot of attention and is directly related to stiffness. There exists a significant positive correlation between GTC and RE (83). Paavolainen and colleagues examined concurrent training effects on running

performance and RE. They examined GCT as a dependent variable in order to explore the mechanisms of neuromuscular adaptation resulting from the training program (93). As well as significant improvements in running performance and RE, the investigators found significant reduction in GCT with no correlating increase in ground reaction forces (93). This study was later supported with further investigation by Paavolainen and colleagues (92) in which 17 male distance runners performed a battery of tests to determine correlations between neuromuscular characteristics, muscular power, and running performance. Paavolainen et al. reported significant correlations between RE and GCT during a 5 km time trial ( $r=0.64, p<0.001$ ).

A similar study conducted years later by Nummela et al. (89), that also demonstrated this relationship between GCT and RE. Nummela et al. (89) studied 25 trained endurance athletes to investigate the interaction between running biomechanics, velocity, and RE. The investigators reported significant correlations between RE and GCT at three of the four velocities tested:  $5.8 \text{ m}\cdot\text{s}^{-1}$  ( $r=0.49, p<0.05$ ),  $6.2 \text{ m}\cdot\text{s}^{-1}$  ( $r=0.44, p<0.05$ ) and  $6.6 \text{ m}\cdot\text{s}^{-1}$  ( $r=0.41, p<0.05$ ). Additionally, both studies found a direct negative correlation between GCT and velocity ( $r=-0.52, p<0.01$  (89);  $r=-0.49, p<0.05$  (92)) during a running time trial, which also correlated significantly with RE ( $r=-0.47, p<0.05$  (89);  $r=0.64, p<0.001$  (92)). These findings are also reflected in the work of Tam et al. (122) who reported a significant correlation between reduced GCT and RE ( $r=0.639, p<0.001$ ) as well as several other studies that have sought to identify the biomechanical characteristics of runners with superior RE (6,19,68,73).

Duty factor (DF) is defined as the ratio of GCT to the duration of the ariel portion of the running stride commonly referred to as the swing phase. The greater the GCT the greater the DF. An analysis of DF aids in the examination of the interaction between leg stiffness, elastic energy utilization, and RE (69). The vast majority of research investigating RE that accounts for GCT

and any aspect of stiffness strongly suggests that RE is negatively correlated to GCT and positively correlated to stiffness, both of which imply a low DF (6,19,68,89,92,122). This consensus is challenged by the recent findings of Lussiana et al. (69). In order to study DF more rigorously and describe the relationship between DF and RE, the researchers collected kinematic and kinetic data on fifty-four trained runners organized into two groups identified as either low- or high-DF. Data was collected at 10, 12, 14, 16 and 18 km·h<sup>-1</sup>. The researchers showed no significant effect of DF on RE ( $p = 0.556$ ) but did show an effect of speed on RE that was dependent upon DF ( $p = 0.025$ ). Lussiana et al. (69) reported a significant decrease in RE ( $-2.3 \pm 2.6\%$  from 10 to 14 km·h<sup>-1</sup>,  $p = 0.008$ ) for the low-DF group but no significant change for the high-DF group ( $1.5 \pm 3.8\%$  from 10 to 14 km·h<sup>-1</sup>,  $p = 0.781$ ). In conjunction with these findings, Lussiana et al. report an increase in step symmetry (ratio of concentric phase of ground contact to eccentric phase of ground contact) with an increase in velocity. This relationship between velocity and step symmetry strongly suggests that the mechanism underlying the observed interaction between velocity and DF, and therefore GCT, is directly related to a nearly isometric action of the muscles of the leg at higher velocities. The observed near-isometric action of the leg muscles would promote greater energy storage and release from the elastic components of the leg including both series elastic components and parallel elastic components (31,69). This observed relationship between velocity and RE is a common theme in research on the mechanisms governing RE (24,52,66,90,92,122).

In conjunction with examination of GCT in relation to RE, investigations of passive stiffness, co-activation of biarticular muscles, fascial stiffness, and fascial contractility have provided insight into the time course of force development and the utilization of elastic energy. Multiple studies have demonstrated a significant negative correlation between the coactivation of

biarticular muscles during the stance phase of running and RE (41,43,63,81,123). The mechanism responsible for this correlation has been identified as an increase in joint stiffness which minimizes hysteresis thus improving the utilization of elastic energy and increasing the rate of force development (23,41,52,61). Aiding in the effect of muscle co-activation, passive stiffness has also been tied to improvements in RE. Increased passive stiffness of the MTUs of the shank has been negatively correlated with cost of running (improved RE) ( $r = -0.69$ ,  $p = 0.01$ ) while similar increase in vastus lateralis stiffness have been correlated with diminished RE (24,79).

In addition to passive stiffness, recent investigations into the contractility of fascia have opened new avenues of investigation of stiffness and elastic energy utilization. Fascial dissection has demonstrated that fascia is a complex matrix of inelastic collagen fibers mixed with elastic fibers (120). Each unique layer of this woven matrix is separated by thin layers of adipocytes which function as lubricant allowing adjacent layers to freely slide over each other (119,120). This organization allows fascia to function in multiple, seemingly contradictory roles: resist traction, adapt to long-term stretch, recoil, and transmit force (119,120). The ability of fascia to transmit force has been demonstrated in animal studies, cadaver studies, and via modelling (48,49,70,97).

Multiple studies have demonstrated the presence of myofibroblasts in human fascia and the ability of this muscle tissue to contract when in the presence of contraction-inducing agents (109–113). These discoveries have interesting implications regarding the nature of elastic energy utilization. Schleip et al. (109) have demonstrated the ability of fascia to actively contract due to the presence of smooth muscle myofibroblasts. While the force generated by these contractions is calculated to be below the threshold for a positive contribution to joint stability/stiffness, it is

strongly suggested that fascial contraction may positively affect passive MTU stiffness and neuromuscular coordination during locomotion (109). Because of the relatively slow time course for the contraction of the myofibroblasts inherent to fascia, this effect requires investigation over a substantially long period of time, minutes to hours in length, and to date has not been investigated under these conditions. The impact that increased fascial stiffness could have on whole-leg stiffness, coordination of the co-contraction of biarticular muscles, passive stiffness, and therefore RE, could be significant and merits further investigation.

Research exploring RE and stiffness has demonstrated a very strong interaction between stiffness, velocity, and RE. Li et al. (66) used a 3D motion capture system (Vicon T40) to measure whole-leg stiffness at three speeds (12, 14, and 16 km h<sup>-1</sup>) and found a significant correlation between RE and velocity. As velocity increased, the correlation between whole-leg stiffness and velocity also increased ( $r = -0.686$ , 12 km h<sup>-1</sup>,  $p < 0.001$  and  $r = -0.761$ , 16 km h<sup>-1</sup>,  $p < 0.001$ ). Dumke et al. (24) measured the stiffness of the triceps surae muscle and tendon separately using the free-oscillation method and noted a significant negative correlation ( $r = -0.69$ ,  $p = 0.01$ ) between muscle stiffness at high velocities (322 m\*min<sup>-1</sup>) and VO<sub>2</sub>. The investigation however reported no significant correlations between muscle stiffness and VO<sub>2</sub> at slower velocities or tendon stiffness and VO<sub>2</sub> at any measured velocity. The researchers note that this difference in their findings compared to those of similar studies is likely due to the chosen method for measuring stiffness and the fact that the stiffness values were obtained at a single joint angle. Giovanelli and colleagues (35) calculated several mechanical parameters in their investigation of the effect of CT in ultra-marathoners. The investigators reported an increase in vertical stiffness (+7.2±1.0%) but a decrease in leg stiffness as a function of speed (-4.6±5.1%) (35). Importantly, the investigators note that the submaximal speeds at which the subjects were

tested were not consistent with typical speeds sustained during a race, relatively slower than average race paces, and therefore may not have elicited the same neuromuscular response in the form of musculotendinous stiffness (35). Jin et al. (51) investigated the modulation of joint stiffness, along with joint work and power, at multiple running speeds (1.8-3.8 m/s). Using inverse dynamics to calculate the specified variables of the ankle, knee and hip joint, the researchers found significant correlations between velocity and joint stiffness. Stiffness at the ankle, knee and hip was positively associated with velocity ( $r=0.96$  ankle,  $r=0.98$  knee,  $r=0.82$  hip,  $p < 0.05$ ). Tam et al. (122) measured several biomechanical and neuromuscular factors in thirty-one trained distance runners in order to identify biomechanical characteristics indicative of more economical distance runners. Among other variables, joint stiffness ( $\text{Nm} \cdot \text{deg}^{-1}$ ) was assessed and compared to the subject's oxygen consumption ( $\text{ml} \cdot \text{kg}^{-1} \cdot \text{km}^{-1}$ ) to evaluate its contribution to RE. The investigators found a significant correlation between oxygen cost of transport and stiffness at the ankle and knee. The experiment revealed a significant positive correlation between ankle stiffness and oxygen consumption ( $r=0.527$ ,  $p=0.007$ ) and a significant negative correlation between knee stiffness and oxygen consumption ( $r=0.384$ ,  $p=0.043$ ).

### *Summary*

Running economy is a crucial component of running performance and is determined by a multitude of physiological, biomechanical, and neuromuscular variables (28,32). The capability of a runner to successfully utilize elastic energy in order to decrease the metabolic cost of running is one of the most important components of RE (20). Stiffness on all levels, whether it is the global vertical stiffness of the runner, whole-leg stiffness quantifying the compliance of the lower limb or musculo-tendinous and aponeurosis stiffness is the most important determinant of

elastic energy utilization. The recent developments in the research of the contractility of fascia and ability of fascia to modulate stiffness in the lower extremities of humans presents new possibilities in the investigation of the underlying mechanisms governing musculotendinous stiffness (109). The new perspective on fascia, coupled with a sophomoric understanding of the role of the iliotibial band and fascia lata in elastic energy utilization (25,26), and the established limitations of current modelling techniques, suggest that further research efforts should be devoted to the role of fascial stiffness and the hip and thigh architecture in RE.

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## **Appendix A: Journal Guidelines to Authors**

*Journal of Strength and Conditioning Research*

Journal guidelines to authors:

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Citation example provided by Journal of Strength and Conditioning Research website:

Hartung, GH, Blancq, RJ, Lally, DA, Krock, LP. Estimation of aerobic capacity from submaximal cycle ergometry in women. *Med Sci Sports Exerc* 27: 452–457, 1995.