

**RELATIONSHIPS BETWEEN MULTIPLE MECHANICAL
STIFFNESS ASSESSMENTS AND PERFORMANCE IN
MIDDLE-DISTANCE RUNNERS**

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Attestation of Authorship

I hereby declare that this submission is my own work and that, to the best of my knowledge and belief, it contains no material previously published or written by another person (except where explicitly defined in the acknowledgements), nor material which to a substantial extent has been submitted for the award of any other degree or diploma of a university or other institution of higher learning.

Signed:

A handwritten signature in black ink, appearing to be 'S. R. ...', written over a light gray rectangular background.

Date: 13 November 2015

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Ethical Approval

Ethical approval for this thesis research was granted by the Auckland University of Technology Ethics Committee (AUTEC) on 21 October 2014, with AUTEC Reference number 14/324 (see *Appendix 3*).

Thesis Overview

This thesis adheres to pathway one, as classified by Auckland University post-graduate thesis structure guidelines (AUT Post Graduate Handbook 2015). The layout of this thesis follows the conventional pathway whereby the document is wholly written. It consists of six chapters with numbered references from each chapter presented at the end of the thesis. Chapter One provides an introduction and overview of the thesis. Chapter Two (Literature Review One) introduces the reader to the concept of stiffness and its common measurement techniques, presenting what each attempts to quantify within human movement. This is complemented with a discussion of the current limitations of such measures. Chapter Three (Literature Review Two), expands on the theory of stiffness outlined in Chapter Two, with a focus on literature that has examined lower-body stiffness as a determinant of middle and long distance running performance. This second review emphasises the current knowledge of the relationships between stiffness and individual differences in running efficiency and performance, with practical recommendations for the use of such measures. Chapter Four describes the methods of the thesis study and includes the adopted experimental design and procedures to measure biomechanical and physiological aspects of running performance. Chapter Five presents the results. Chapter Six is the discussion and provides an evaluation of the study findings including limitations, applications of findings and areas of potential future research in this area.

List of Abbreviations and Definitions

- 5-H** **Five consecutive hopping test:** as performed on the left and right limb separately, and also bilaterally.
- CoM** **Centre of mass:** a point representing the midpoint of the body mass.
- E_{run}** **Energy cost to run:** expressed as a caloric cost per kilogram of body weight to run a kilometre.
- GRF** **Ground reaction force:** typically referring to the vertical forces, unless otherwise specified.
- k_{AT}** **Achilles tendon stiffness:** based on the displacement of the tendon of which segments can be imaged (*in-vivo*) alongside a recording of the contributing forces of both the agonists and antagonists.
- k_{leg}** **Leg stiffness:** based on the spring-mass model, it represents the resistance to change in leg length occurring from initial ground contact to the bottom of stance; a measure of individual leg springs.
- k_{vert}** **Vertical stiffness:** based on the spring-mass model, it represents how the human body, as a whole, responds to the ground reaction force applied to the system where there is no horizontal displacement (i.e. hopping or jumping).
- L_0** **Resting standing leg length:** measured as the height of the greater trochanter from the floor when standing without shoes.
- MD** **Middle-distance:** running events comprising the 800 m to 5000 m in athletics.
- MVIC** **Maximal voluntary isometric contraction:** in this study performed during plantarflexion in an isokinetic dynamometer.
- PF** **Plantarflexion:** of the triceps surae and Achilles tendon MTU.
- PLYO** **Plyometric training:** using explosive or ballistic training movements.
- RE** **Running economy:** reflects the energy demand of submaximal running. Those with good economy will use less oxygen than those with poor economy at the same steady-state speed or intensity.
- S_f** **Step frequency:** the number of foot contacts per second.
- sL** **Step length:** distance (m) between toe to toe in subsequent foot contacts.
- SMM** **Spring-mass model:** a mechanical model following the principle of Hooke's law for a linear spring.

Tc Contact time: time interval when the foot is in contact with the ground during the stance phase of gait.

Tf Flight time: time interval from when the foot leaves the ground, to the time of next foot contact on the opposite limb

v_{\max} Maximal running or sprinting velocity: in this study performed overground in racing spikes

ABSTRACT

Background: The ability to sustain high running speeds during competitive races is a key determinant of success in elite middle-distance (MD) runners. Rapid and economical movements are, therefore, highly valued characteristics in their ability to determine performance. It is suggested lower-body stiffness may serve as a valuable measure of a runner's compressibility and energy utilization during their ground contact interactions at both maximal and submaximal speeds. Unfortunately, evidence concerning the relationship between stiffness and running performance is convoluted, and a lack of consensus regarding optimal stiffness assessment methods leave a degree of confusion for practitioners and researchers alike. **Aim:** To quantify the levels of lower-body stiffness in highly-trained male MD runners and determine whether stiffness is a determinant of running economy (RE) and maximal velocity (v_{\max}). Further, the study aimed to determine the relationships between multiple levels of stiffness measures calculated from both laboratory and field settings. **Method:** Using a cross-sectional analysis, eleven highly-trained male MD runners (Mean \pm SD: age: 20.0 ± 2.9 yr; body mass: 68.2 ± 6.7 kg; $\dot{V}O_{2\text{peak}}$: 67.6 ± 3.8 ml \cdot kg $^{-1}\cdot$ min $^{-1}$; 1500 m personal best $4:02 \pm 0:06$ min:s) performed maximal overground sprints and a submaximal treadmill test to determine RE followed by a ramp protocol to elicit $\dot{V}O_{2\text{peak}}$. Multiple assessments of stiffness were conducted based on the spring-mass model (SMM), including leg (k_{leg}) and vertical stiffness (k_{vert}) during running and hopping respectively. Additionally, the Achilles tendon stiffness (k_{AT}) was estimated using ultrasound during maximal isometric ankle plantarflexion. **Results:** Stiffness values were comparable with previous literature on trained runners. There was a *very large* negative correlation ($r = -0.70$) between RE and k_{AT} . RE had *large* and *moderate* negative relationships ($r = -0.60, -0.46$) with k_{vert} during the maximal hopping tasks on the right and left limbs respectively. When k_{leg} was calculated during submaximal running, the association with RE was *unclear*. In addition, *large* positive correlations were detected between k_{AT} and v_{\max} ($r = 0.52$) and between k_{leg} and v_{\max} during sprinting ($r = 0.59$). Finally, examining the association between stiffness methods, k_{vert} during maximal unilateral hopping (right leg) and k_{leg} during sprinting has an *extremely large* ($r = 0.92$) relationship. Measures of k_{leg} during sprinting also held *very large* ($r = 0.73$) relationships to unilateral k_{vert} and k_{AT} . **Conclusion:** This study demonstrated a clear association between both global and component stiffness measures and a faster and more economical running performance in trained runners. These findings in the context of MD runners of this calibre are unique. Although k_{AT} had

the strongest relationship with RE, the use of more practical methods of measuring stiffness is recommended. Namely, stiffness expressions utilizing the SMM in maximum velocity sprinting (primarily) and maximal unilateral repeated hops (secondly) appear to be useful surrogates for field-based assessment of stiffness related to MD running performance. Agreement between the SMM assessments and k_{AT} , highlighted the contribution of the muscle-tendon structures of the lower-limb to models of human running. Researchers and practitioners should therefore consider SMM stiffness assessments from a range of tasks to profile MD running performance.

CHAPTER ONE

INTRODUCTION

1.1. Background

Running is a primal yet complex human movement, involving the conversion of muscular power to highly tuned motor patterns utilising all major structures within the body. Since the dawn of the modern Olympic era, optimizing the performance of this seemingly simplistic act has been widely researched and debated by scientists and coaches from primarily physiological and biomechanical perspectives.

Middle-distance (MD) running encompasses distances of 800 m to 5000 m, with elite competitors required to possess the ability to move at a high velocity while keeping actions as efficient as possible [5]. During competition, mean race velocities of international male MD runners reach $\sim 7.65 \text{ m}\cdot\text{s}^{-1}$ [5], with the ability to sustain high velocities at supra-maximal aerobic speeds being critical to race success [6]. A significant aerobic capacity ($\dot{V}O_{2\text{peak}}$), alongside the anaerobic component of MD events, create a requirement for both a high velocity at $\dot{V}O_{2\text{peak}}$ denoted as $v\text{-}\dot{V}O_{2\text{peak}}$, which is a measure representing the relationship between $\dot{V}O_{2\text{peak}}$ and running economy (RE) [6, 7], and maximal sprinting velocity (v_{max}). With the final lap velocity often determining major championship outcomes [8], v_{max} appears increasingly imperative to success.

Training methods for $\dot{V}O_{2\text{peak}}$ and RE have traditionally focused on modifying (often increasing) each of, or collectively, training intensity, duration and frequency in order to achieve positive adaptations [9-11]. Although RE is widely acknowledged to be informed, to a degree, from physiological determinants, it is also influenced by underlying modifiable and non-modifiable mechanical factors [12]. Successful RE adaptation and subsequently performance improvements are thought possible via mechanical alterations of the neuromuscular system from running based [13] and non-running training interventions [14-17]. A frequently cited mechanism underpinning neuromuscular adaptation to training is mechanical stiffness, a measure which attempts to quantify the production and the resistance of movement [1, 18, 19].

Stiffness is defined as the relationship between a given force and deformation or compression of an object or body [18]. Lower-body stiffness can be used to describe individual muscle fibres or tendons, also to model resistance to compression across the whole body [20]. In research and practical settings, stiffness has been assessed through a hierarchy of systems, each level adding more detailed information about human movement. During running and jumping, the musculoskeletal system responds to the impact of each landing through compression of the leg or legs, modelled as ‘springs’ [1, 4]. These elastic structures, or ‘springs’ of the lower body, store kinetic energy with each ground contact during gait and lengthen eccentrically [2, 21]. In the subsequent propulsive phase, a portion of this stored elastic energy contributes to a runner's forward movement [22]. As greater forces are imparted to the body with increasing velocity, a runner requires a greater amount of neuromuscular control of the limbs in space, in conjunction with efficient and powerful propulsion [18]. Therefore, stiffness provides a representation of a runner’s lower-body capacities of both efficiency and power during this eccentric-concentric action, an action commonly known as the stretch-shortening cycle (SSC) [23].

The stiffness principle has been applied to model the body as a linear ‘spring’, as well as to examine specific and isolated component structures of the limbs. Global stiffness, based on a simple ‘spring-mass model’ (SMM), describes a mass (the body) atop a massless spring (the leg) following the principle of Hooke’s law for a linear-spring [1, 24]. Global stiffness comprises three commonly discussed classifications: vertical (k_{vert}), leg (k_{leg}) and joint (k_{joint}) stiffness, each incorporating a greater number of musculoskeletal and neural components. These three descriptions contribute to the measurement of the body’s resistance to displacement after the application of ground reaction forces (GRF’s) [25] while each successive level of measurement requires a more extensive data collection and analysis process. Isolated or component measures of stiffness examine various muscle-tendon units (MTUs) [26] and attempt to understand contributions of a specific structure (typically *in-vivo*) which sits within the ‘leg-spring’. The use of ultrasound enables imaging of MTUs and measurement of length changes during a maximal voluntary isometric contraction (MVIC) to quantify the material and mechanical properties [27]. For example, the lower-limb components of the Triceps-surae (TS) and Achilles tendon (AT) return a large amount of this stored elastic energy in locomotion [28] and has been observed to reduce the metabolic demand of lower-limb muscles and to potentially improve RE [15, 16, 29].

By association, several studies [29-33], but not all [34], have suggested that a ‘stiffer’ leg-spring is theoretically advantageous to efficient distance running performance when stiffness is measured either as k_{vert} , [31], k_{leg} [29] or k_{MTU} [32]. Similarly, a relationship between stiffness and RE is evident in longitudinal studies [14-17] that explored numerous non-running conditioning modalities and their subsequent effects on stiffness, RE and race performance. Spurrs et al. [16] stated that moderately-trained runners can elicit performance improvements from strength training in as little as six weeks and changes in stiffness may partially explain such improvements. Collectively, these findings suggest stiffness may offer an explanation of inter-individual differences in running efficiency [35, 36], and maximal velocity (v_{max}) [20, 37]. However, it remains unknown whether an optimal level of stiffness is required for a given level of performance [18, 19]. Previous associations between v_{max} and stiffness have only been reported in sprinters [37, 38] and recreational athletes [39], with little known about the relationship in highly-trained MD runners. Due to the large array of stiffness assessments adopted across a spectrum of running abilities, there is a need to further understand the extent to which stiffness is associated with key physiological and biomechanical variables in highly-trained MD athletes. Additionally, the link between the global and component-based measures of stiffness remains elusive, with only one study [40] reporting stiffness using different methods, that revealed no relationship between hopping k_{vert} and ankle stiffness from an oscillation task, in female recreational athletes. There is a lack of published studies examining global and component stiffness assessments alongside multiple determinants of performance in the highly-trained MD running population. Further, no studies appear to have quantified the degree of agreement between stiffness measures from laboratory and field settings. By addressing this void, this thesis will aid in athlete profiling, assessment selection and in guiding future research in MD running events.

1.2. Study Aims

1. To quantify the levels of lower-body stiffness in highly-trained male MD runners and determine whether stiffness is one determinant of running economy and maximal velocity in highly-trained male middle-distance runners.
2. To determine the relationships between multiple levels of stiffness measures calculated from both laboratory and field settings.

1.3. Significance of the Study

Practitioners who work in both the assessment and implementation of endurance training programmes are charged with quantifying and enhancing the factors that limit performance. Stiffness measures may be a potential addition to the mechanical and physiological profiling and monitoring of training adaptations. However, current conflicting evidence prevents practitioners having the confidence to make meaningful inferences from a practical measure of stiffness. Therefore, the results of this investigation will add to the understanding of how global and component stiffness assessment techniques interrelate and how each may link to physiological and mechanical performance measures in highly-trained MD runners. The establishment of an athlete profile that addresses both mechanical and metabolic performance determinants will assist in enhanced individualised & quantifiable training approaches.

CHAPTER TWO

LITERATURE REVIEW ONE

Determining lower-body mechanical stiffness: A review of models and methods

2.1. Introduction

Interest in biomechanical lower-body stiffness has grown in popularity among sports science communities, as it may impact on both running performance and on injury prevention [18]. Stiffness is defined simply as: *the relationship between the deformation of a body and a given force*. Stiffness has been used to model the human body as it reacts with the ground during landing, in addition to describing physical properties of individual muscle-tendon units (MTUs). Given this definition, stiffness can represent a massless spring, where force (F) exerted to deform a body, is related to a proportionality constant (k) and the distance (x) the body is deformed in the opposing direction, as described by Hooke's Law of $F = kx$ [18, 28]. Rearranged as: $k = F/x$, to determine the spring constant, this relationship describes the stiffness of an ideal spring-mass model (SMM) and is used to describe human jumping and locomotion [1, 4].

In a classical view of running mechanics, a runner's response to the impacts of landing are based on the displacement of their centre of mass (CoM), achieved through the compression of the 'leg springs' originally termed the 'bouncing gait' [1, 4]. In this simplified model, when in the stance phase the leg behaves as a massless mechanical spring loaded by the body during each foot strike [2, 21]. During the eccentric phase of ground contact, an athlete stores mechanical energy in the elastic tissues of the leg [41]. In the subsequent concentric recovery period, stored elastic energy contributes to the body's next movement, reducing overall energy expenditure [17, 42]. An optimal level of stiffness reportedly exists for an individual during a given movement that may in part be determined by the stretch-shortening cycle (SSC) for power output and efficacy [18, 43]. Stiffness has been related to several key running performance measures including running economy (RE) [15, 29] and maximal running velocity (v_{\max}) [20].

Since its inception, the validity of this SMM has often been questioned. That such a simple model attempts to describe a multitude of influences on human locomotion in a single measure is debatable [28]. Compression of the 'leg-spring' which occurs via

rotation of the hip, knee and ankle joints (each holding multiple degrees of freedom), is controlled by surrounding muscles, ligaments and tendons [1, 44]. Clark and Wayand [39] recently summarised several key assumptions when in the SMM, including; i) the ground reaction force (GRF) versus time waveform represents the shape of a half-sine wave, ii) the displacement of the body's CoM during the compression and rebound portions of ground contact are symmetrical, and iii) the F_{max} will occur in mid-stance when the CoM reaches its lowest position. Global values of stiffness are made from kinematic, kinetic and spatiotemporal variables [45, 46]. Measures associated with producing a global stiffness value, based on the SMM, could more correctly be termed 'quasi-stiffness,' as they result from actions of both elastic and contractile components, along with the several mechanical assumptions outlined above [29, 39]. Stiffness in the component sense, examines a given muscle-tendon unit (MTU) *in vivo*, together with the associated physiological and mechanical structures [26]. A variety of techniques for modelling and measuring stiffness are often dictated by task constraints and the equipment available [2]. Thus, the full link between modelled and structural stiffness remains elusive.

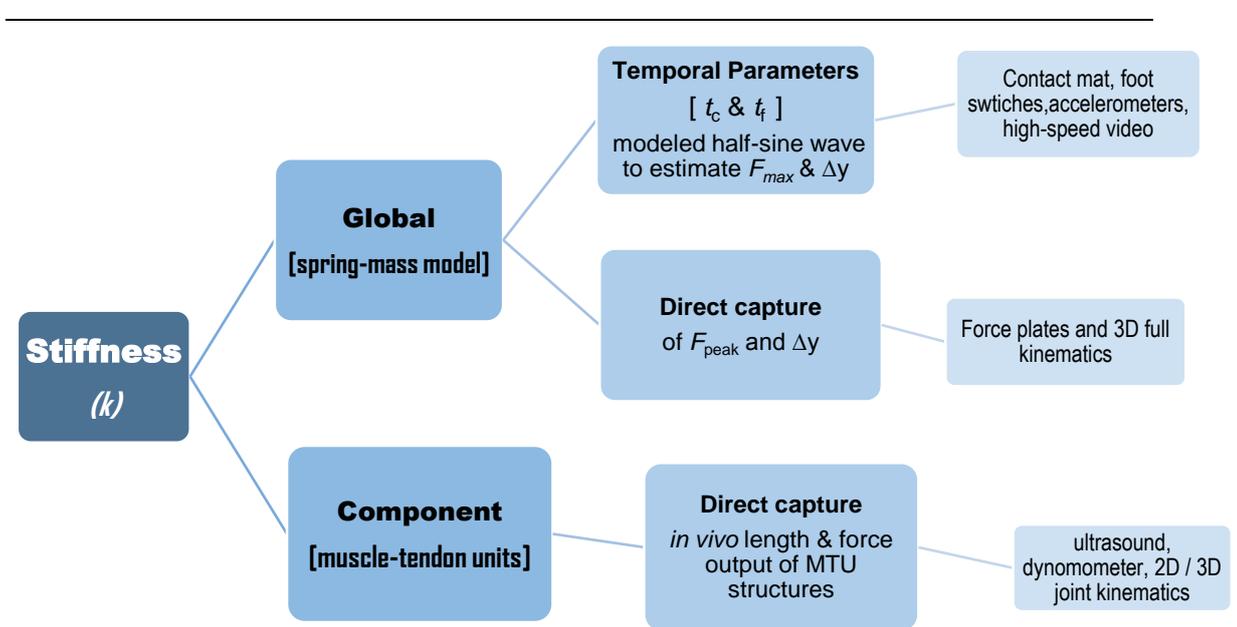
2.1.1. Aims of the review

The aim of this review is to collate and summarise both global and component forms of biomechanical stiffness with a focus on discussing methodological differences and limitations within this body of research. This will allow researchers to make informed decisions in future studies around measures of stiffness as an influencing factor in running performance.

2.2. Stiffness measurement classification

Lower-body stiffness methods are hierarchical, with each measure adding more detailed information about human movement [21, 47]. These levels of measurement can be categorised from modelling the whole body as a linear spring, through to evaluating a specific tendon. Each stiffness measure is intrinsically linked, yet more specific at each descending level [48]. Despite its mechanical simplicity, the SMM has provided accurate predictions of the vertical force vs. time waveforms represented during slow and intermediate running speeds [1, 39, 49, 50]. To assess force and time measures, data is commonly obtained via one of two techniques: i) directly or dynamically through the use of force transducers (either in-ground or mounted below a treadmill), reported as the gold

standard [2, 21, 50]; or ii) indirectly calculated from temporal measures (e.g. contact and flight time) [2], obtained using equipment such as contact mats, accelerometers or high-speed cameras [51]. The latter method is frequently employed when obtaining direct force measurements is impractical, for example in applied sports settings. These factors are discussed in part 2.5 of this review. Alternatively, component stiffness measures which involve direct measurement of specific muscle and tendon units, typically use isometric (ISO) muscle contractions to make *in vivo* structural and mechanical assumptions. Such measures can be achieved through a combined use of ultrasonography and dynamometry equipment in laboratory settings [52]. Methodological considerations for both the global and component measures are discussed in detail in parts 2.3 and 2.4 respectively and summarised in *Figure 2-1*.

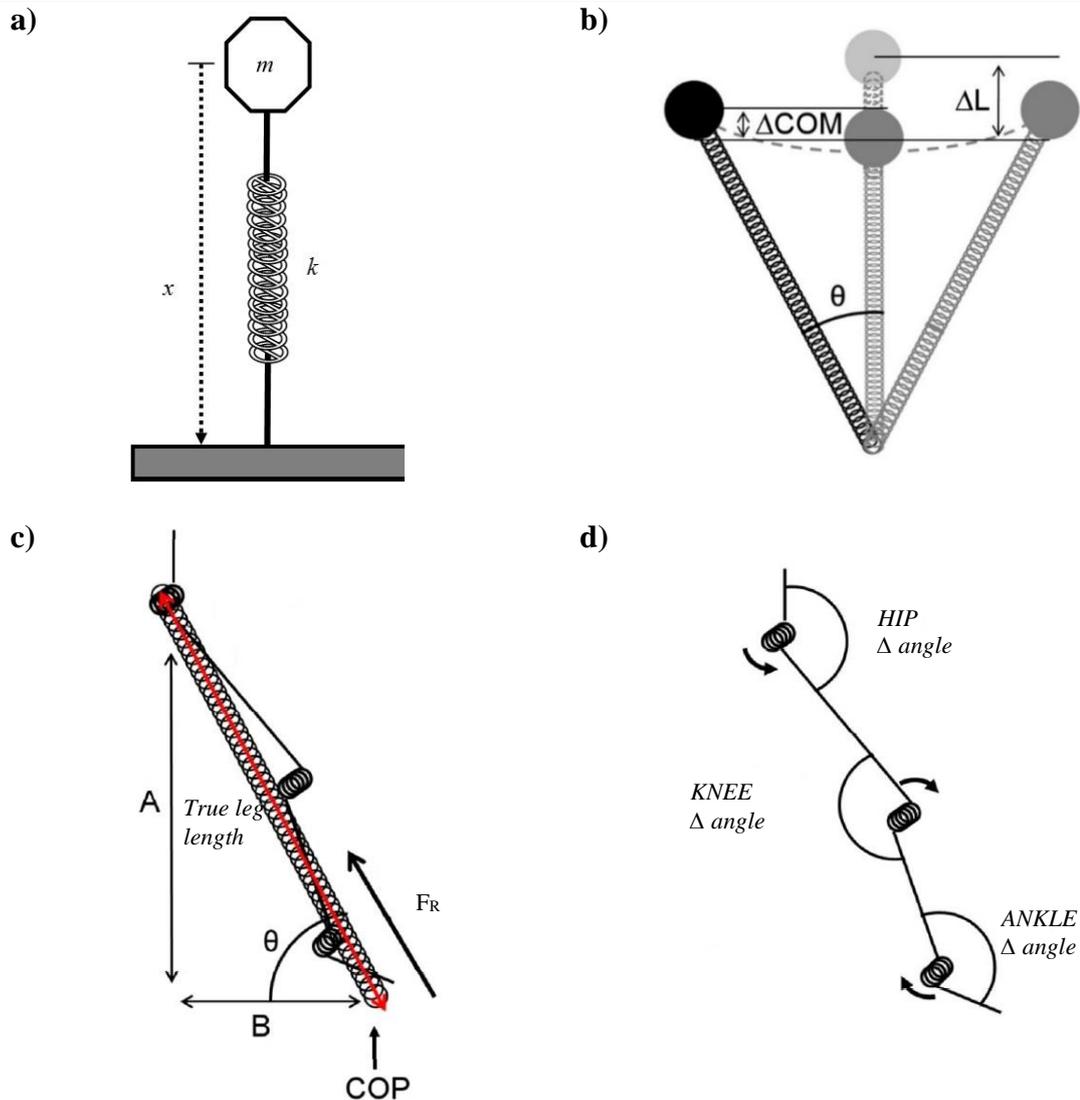


F_{max} = peak vertical force, Δy = centre of mass displacement, t_c = contact time, t_f = flight time

Figure 2-1. Summary of global and component biomechanical stiffness measures with key methodological requirements.

2.3. Global measures for modelling lower-body stiffness

Global lower-body stiffness refers to the response of a multiple-component system to loading and often incorporates both passive and active elements. In relation to human running gait, these components are typically divided into three categories: vertical, leg, and joint stiffness – outlined below and illustrated in *Figure 2-2*, with calculations of each presented in *Table 2-1*.



Model a) k = spring constant, m = mass, x , vertical displacement, **Model b)** ΔCOM = maximal displacement of centre of mass, ΔL = change in leg length during compression, θ = angle of leg swing, **Model c)** COP = centre of pressure, A = position of the greater trochanter from the ground using 3D analysis, B = distance along the x-axis between the greater trochanter and the COP, F_R = resultant of the vertical & horizontal force, **Model d)** arrow rotation = Δ joint moment

Figure 2-2. Schematic representations of lower-body biomechanical stiffness following the spring-mass model for:
 a) k_{vert} and b) k_{leg} based on model by McMahon and Cheng's [1];
 c) Coleman et al.'s [2] 'True' leg stiffness model; d) Joint stiffness model [3, 4].

Source: Lorimer [48]

2.3.1. Vertical stiffness

Vertical stiffness describes the resistance of the body to movements of the CoM occurring in the vertical plane [18]. When the resulting forces are measured via force transducers, vertical stiffness ($k_{\text{vert/dynamic}}$) is calculated as the peak vertical GRF divided by maximal vertical displacement (Δy) of the CoM during ground contact. The Δy determined from double integration of the vertical force curve, was first described by Cavagna [24]. This process for calculating $k_{\text{vert/dynamic}}$ (Table 2-1, Eq. 1) holds the assumption that the position of the CoM is similar at take-off and on landing, resulting in an integration constant equal to zero [18]. Additionally, quantifying vertical stiffness using temporal measures to estimate force ($k_{\text{vert/time}}$) employs a sine-wave method, described and validated previously by Morin et al. [21]. Notably, this model for measuring stiffness does not require expensive force plate technology, nor is it confined to measurement in a laboratory environment.

2.3.2. Leg stiffness

The concept of leg stiffness (k_{leg}), is based on the model of k_{vert} of the leg-spring when the CoM is not directly over the initial ground contact point, as in running [18]. The angle of leg swing (θ_{leg}) is accounted for when calculating the amount of leg compression or length change (ΔL) as it sweeps through the arc during stance. Additional variables required for this value of stiffness include resting leg length (L_0), horizontal velocity (v), contact time (t_c) and peak vertical GRF (Table 2-1, Eq. 3& 4). There is confusion in the literature between nomenclature of vertical and leg stiffness, as these terms have often been used interchangeably. In essence, both model the leg's stiffness following the same assumptions, differing only in the method for calculating ΔCoM or ΔL . In k_{leg} , the θ_{leg} is taken into account from known quantities of v and L_0 . If the body is moving only in the vertical direction (as in hopping), then θ_{leg} becomes zero and k_{vert} is equal to k_{leg} [18]. To ensure clarity is maintained between each model's calculations, vertical and leg stiffness will be referred to as assessed via hopping (vertical) or running (leg) tasks respectively.

Similar to the dynamic and time-based measurements of k_{vert} , using the model proposed by Morin et al. [21], a measure of $k_{\text{leg/time}}$ can also be obtained from t_c and t_f to estimate F_{max} (Table 2-1, Eq. 4). The mathematical modelling to obtain estimates of $k_{\text{leg/time}}$ use double integration of the vertical acceleration of the body to estimate the compression of the leg spring and to calculate its ratio to peak vertical GRF [21, 53-55]. Utilizing this field-based method of $k_{\text{leg/time}}$ does not require the runner to 'target' their foot strike(s)

over a force plate, which potently alters their gait characteristics [2, 4]. Alternatively, capturing only t_c and t_f allows an estimate of stiffness during a sport specific task. A recent review [25] highlighted several different methods for estimating ΔL (peak displacement of the leg, from the initial leg length). Estimation of ΔL from the dynamic force versus time base variables has produced differing values of k_{leg} when compared to a recently proposed ‘true’ measure of ΔL through three-dimensional (3D) modelling [2]. Additional dynamic estimations of ΔL through 3D kinematics and kinetics have only been utilized in several studies for running measures [56, 57] and are summarised in *Table 2-2*. Further differences between $k_{leg/dynamic}$ and $k_{leg/time}$, are discussed in part 2.5 of this review.

2.3.3. Joint stiffness

Running requires the hip, knee and ankle joints to act synergistically as a part of the human body’s kinetic chain to control landing during gait [48]. Therefore, it seems intuitive that leg and vertical stiffness emerge from local elasticity established by joint torques or moments [58]. As these joints absorb GRFs at impact, joint stiffness (k_{joint}) acts as a modulator to the various degrees of rotation and is important in overall lower-body stiffness. Individual k_{joint} values describe the resistance to this rotation and are defined as: *the change in a joint moment, divided by the change in joint angle* (*Table 2-1*, Eq 5). Eccentric strength and antagonist muscle activity play a role in the degree of k_{joint} , in which deficiencies contribute to a limited ability to control rates of joint rotation (i.e. a lack of stiffness) [54, 59], which may be linked to injury [43]. Unsurprisingly, a greater level of detail about the mechanics of landing can be obtained from k_{joint} compared to the prior two levels of global stiffness. However, research surrounding the individual contributions of each joint with the improvement of running performance is debatable due to a paucity of literature involving trained populations [48].

Table 2-1. Summary of ‘Global’ biomechanical stiffness model equations.

Source: Adapted from Lorimer [48]

Classification	Eq.	Model (original reference)	Calculation	Key variables (<i>equipment</i>)
<i>Vertical stiffness</i>	(1)	Dynamic vertical stiffness (McMahon and Cheng [1])	$k_{\text{vert/dynamic}} = \frac{F_{\text{max}}}{\Delta y}$	Vertical GRF (<i>force plate</i>)
	(2)	Time/temporal vertical stiffness (Morin et al. [21])	$k_{\text{vert/time}} = \frac{F_{\text{max}}}{\Delta y}$ $F_{\text{max}} = mg \frac{\pi}{2} \left(\frac{t_f}{t_c} + 1 \right) \& \quad \Delta y = -\frac{F_{\text{max}} t_c^2}{m \pi^2} + g \frac{t_c^2}{8}$	Contact & flight time (<i>contact mat, 2D high-speed video</i>)
<i>Leg stiffness</i>	(3)	Dynamic leg stiffness (McMahon and Cheng [1])	$k_{\text{leg/dynamic}} = \frac{F_{\text{max}}}{\Delta L}$ $\Delta L = \Delta y + L_0(1 - \cos\theta) \& \quad \theta = \sin^{-1} \left(\frac{vt_c}{2L_0} \right)$	Vertical GRF (<i>force plate/instrumented treadmill</i>) v (<i>radar/timing lights/treadmill</i>) L_0 standing leg length
	(4)	Time estimated Leg Stiffness (Morin et al. [21])	$k_{\text{leg/time}} = \frac{F_{\text{max}}}{\Delta L}$ $F_{\text{max}} = mg \frac{\pi}{2} \left(\frac{t_f}{t_c} + 1 \right)$ $\Delta L = L_0 - \sqrt{L_0^2 - \left(\frac{vt_c}{2} \right)^2} + \Delta y$ $\Delta y = -\frac{F_{\text{max}} t_c^2}{m \pi^2} + g \frac{t_c^2}{8}$	Contact & flight time (<i>contact mat, 2D high-speed video</i>) v (<i>radar/timing lights/treadmill</i>) L_0 standing leg length
<i>Joint stiffness</i>	(5)	Joint Stiffness (Farley et al. [44])	$k_{\text{joint}} = \frac{\Delta M}{\Delta \theta}$	3D kinematics and kinetics (<i>force plate / instrumented treadmill, 3D motion capture</i>)

F_{max} = peak vertical force, Δy = centre of mass displacement from double integration of F_{max} , m = subject mass, t_c = contact time, t_f = flight time, g = acceleration due to gravity, ΔL = change in leg length, L_0 = standing trochanterian height, θ = angle of leg swing, v = horizontal velocity, ΔM = change in joint moment, $\Delta \theta$ = change in joint angle, GRF = ground reaction force, 2D = two dimensional, 3D = three dimensional.

Table 2-2. Dynamic measurements of leg compression in stiffness assessments.

Study (y)	Sample (N)	Leg length	Force calculation	Running Task
Coleman et al. [2]	19 M; highly-trained runners	'true' Δ leg length; HJC to the COP of foot during ground contact	Resultant F_{\max} in the direction of leg compression	Overground, 60 m runway, 1 in-ground force plate at 30m
Grimmer et al. [56]	11 M; athletes (non-specialist runners)	Δy of the HJC relative to ball of the foot during ground contact	Vertical F_{\max}	Overground, 17 m long track, 3 in-ground force plates
Stafilidis and Arampatzis [57]	10 (M) well-trained sprinters	Δy of HJC relative to point of force application from touchdown until F_{\max}	Vertical F_{\max}	Overground, 60 m running track, 4 in-ground force plates

M = male, COP = centre of pressure, F_{\max} = peak ground reaction force, Δy = change in vertical displacement, HJC = hip joint centre (trochanter major)

2.3.4 Methodological considerations for measuring global stiffness

2.3.4.1. Hopping Measures

For time-efficient assessment of stiffness, researchers employ hopping as a substitute for running if testing space or resources are limited [4]. Different forms of hopping techniques have been performed (maintaining straight or bent knee); thus it may not be appropriate to draw comparisons across different studies. Currently, limited research exists on the direct associations between stiffness estimates from hopping and running performance measures within the same well-trained athletes. Chelly and Denis [37] reported that sprinter's k_{vert} was significantly related to their track and treadmill running v_{max} ($r = 0.68$ and 0.73 ; $P < 0.05$, respectively). A comprehensive study by Lorimer [48] in well-trained triathletes assessed all three levels of global stiffness. Lorimer's work revealed hopping with a straight knee (as opposed to bent) in time with a metronome (2.2 Hz), provided similar knee stiffness ($k_{\text{joint/knee}}$) values in relation to $k_{\text{joint/knee}}$ during submaximal running when obtained using 3D motion capture. The reported $k_{\text{joint/ankle}}$ was lower in both hopping conditions (straight and bent leg) when compared to measures of running. Lorimer attributed this to the notion that while running, there are greater flexions and moment arm changes at the knee joint relative to the hip and ankle joints. It can be surmised that k_{leg} is largely modulated by the knee, allowing adaptation to the environment/surface following ground contact [48, 60, 61]. By contrast, k_{vert} while

hopping was largely determined by $k_{\text{joint/ankle}}$ [44], suggesting a greater understanding of this joint and its structural components would be insightful.

Hopping studies have also used a variety of frequencies to measure stiffness [46]. Performing hopping constrained to a frequency determined by a metronome, may not align with the subject's preferred rate, thus altering a natural movement oscillation where their stiffness may be optimised [46, 62]. As the SMM is informed by contact time, imposed hopping and step frequencies have shown to affect stiffness [46]. Hopping rates that more closely mimic the simple spring-mass system have been suggested to be ≥ 2.2 Hz [48, 63]. Additionally, previous literature suggested a self-selected frequency may minimise energy expenditure during human movement [64, 65]. Further, there appears to be gender-based differences in self-selected hopping frequencies that may or may not alter stiffness measures [62]. Arampatzis et al., [66] described how verbal instructions were important when conducting hopping tests for stiffness. Verbal instructions influenced contact time and subsequent calculations when aimed at maximal height versus maximal take-off velocity. Arampatzis et al. also found an optimal k_{leg} existed that maximized an individual's mechanical power during the propulsive phase of drop jumps of various heights as moderated by contact times. It should be noted, however, that hopping produces a small degree of horizontal movement, as there is a shifting of a subject's take-off and landing position during hopping tasks [48]. The horizontal movement is assumed to be negligible in models for k_{vert} .

2.3.4.2. Running measures

Two key variables previously reported that influence running measures of stiffness are velocity and surface. In a review of stiffness methods, Brughelli and Cronin [60] observed that in three studies employing an identical calculation of k_{joint} [67-69], an increasing running speed elicited an increased $k_{\text{joint/knee}}$, while $k_{\text{joint/ankle}}$ remained constant. These $k_{\text{joint/knee}}$ alternations supported the knee structures as being key modulators of overall stiffness during running and highlighted the importance of its measurement. Similarly, as for hopping, there are also differences in stiffness values between running at self-selected velocities and in setting an absolute velocity for inter-individual comparison [20]. The choice of the task should be considered, particularly concerning investigations between genders and different levels of athletes.

Researchers frequently use measures from treadmill running to make inferences for overground running (training and competition). Several biomechanical differences between treadmill running and overground running have previously been identified [70, 71]. These differences have the potential to impact measures of stiffness using computational models. Relating specifically to stiffness, treadmill running leads to a more upright posture, with a shorter step length (S_L), higher step frequency (S_f) and a flatter foot strike than overground running [70]. Since S_f is a composite parameter of both contact and flight time [55], measures of S_L and S_f are linked with an increased k_{leg} [48, 55, 72]. Furthermore, running over-ground requires the CoM to move over the centre of pressure (CoP) during stance, while the opposite is true for treadmill running, whereby the CoP moves under the CoM. This treadmill effect results in a smaller CoP-to-hip distance throughout the stance phase [48]. Thus, if joint angles are consistent between running conditions, the shorter CoP distance produces a longer effective leg, making the ΔL smaller for treadmill running. Despite these minor differences, treadmill running is advantageous in its capacity to allow for a large number of steps at a fixed velocity to be used to calculate average stiffness for a given period or distance. Using treadmills have proved useful in determining stiffness changes resulting from fatiguing protocols [73, 74].

Additional issues complicating both hopping and running stiffness have arisen from the paucity of literature that examined sport specific highly-trained athletes, together with the applicability of findings from other populations. Even within the overarching sport of distance running, the requirements of the specific events (i.e. middle versus long distance races) potentially dictate the degree of stiffness required for optimal performance, due to the difference in mean competition velocity [5]. Therefore, further study that compares subjects of different age, gender and training status across the same protocol, may provide greater understanding to how measures differ within and between sporting populations.

2.4. Component measures for modelling lower-body stiffness

The models of stiffness discussed thus far have examined the relationships between kinetics and kinematics, only inferring specific muscle–tendon behaviours during the SSC movements. However, component stiffness involves *in vivo* measures of the MTU via real-time ultrasound imaging. Mechanical properties of MTUs are typically calculated while a ramped isometric contraction is performed on a force plate or against a dynamometer. Such components of the leg have been proposed to effect vertical, leg and

joint stiffness during running [60]. Component stiffness, is the change in force applied by a tendon per change in tendon length [75], and contributes to an improved understanding of the physical and mechanical properties of human movements. To the best of my knowledge, only a handful of studies have attempted to record *in vivo* tendon measures during running [76-79]. Unfortunately, there has been little research conducted on this method. The scope of the current review therefore, focuses on component stiffness from isometric assessments of the tendon. Some studies [16, 30] have inferred tendon properties without imaging, through modelling the lower limbs of runners as dampened springs during a static load stabilisation task. These methods are an indirect assessment of the elastic properties of limb components and are discussed in part 2.4.2. Component stiffness holds important links to the understanding of mechanisms responsible for chronic or acute alterations in stiffness. These alterations have been linked to positive changes in sports performance following various loading protocols [75, 80-82].

2.4.1 Measurement of the muscle-tendon unit

In the lower-limb, the triceps surae (TS) muscle group encompasses the medial and lateral gastrocnemius and the soleus, which join distally to form the Achilles tendon (AT). These structures make up the MTU of the lower limb. The AT plays a key role in energy production and forward propulsion in running [79]. This connective tissue structure is an excellent accumulator of elastic mechanical energy and is highly sensitive to routine mechanical loading [28]. Force regularly applied to a tendon has been reported to increase its stiffness and material properties [83] and the AT reportedly resists up 12.5 times body weight during peak loading in running [84]. During the stance phase of running gait, the AT lengthens allowing the TS muscle fascicles to operate at speeds favourable for efficient energy production [80].

Tendon stiffness (k_T) variables are obtained when performing a maximal voluntary isometric contraction (MVIC), for example, ankle plantarflexion (PF) using an isokinetic dynamometer. However, performing this task in the optimal position to measure and isolate force from the lower limb components, is debatable. To allow the entire TS to contribute to a MVIC, a study by Joseph et al. [81] set participants with their knee at full extension and ankle at 90° while seated in the dynamometer chair. The protocol used by Joseph et al. was based on work by Miaki, Someya and Tachino [85], who reported that electromyography (EMG) activity of the soleus was not statistically different ($P < 0.05$) across three knee flexion angles. However, the gastrocnemius did show significantly

greater activity with a 130° knee position and the ankle neutral (90°). Literature would suggest that more the recent technique of positioning the participant prone is more practical while the foot is secured in the dynamometer attachment [17, 32, 36, 86]. Positioning participants and the ultrasound probe securely allows capture of the displacement of the muscle–tendon junction or aponeurosis for accurate and estimation of tendon elongation [32, 83].

2.4.2. Measuring oscillation in the lower-leg springs

The oscillation or sinusoidal perturbation technique is an indirect measure of the properties of the muscle-tendon components of the lower-limb in a closed kinetic chain movement. Several researchers have attempted quantifying stiffness in this way, based on the assumption that the contractile and non-contractile elements of the limbs behave like a damped spring system [87, 88]. Spurrs, Murphy and Watsford [16], presented this technique that had participants perform a MVIC on an instrumented seated calf raise machine (*Figure 2-3*) with a load cell placed under a rigid footplate. Measurement of the maximal force produced by a single limb was acquired via a fixed instrumented knee pad. When performing the MVIC, a chain (*Figure 2-3*, dashed line), held the machine in a fixed position. Following the MVIC, the chain was removed, and a load added to the apparatus, above the knee. In subsequent trials, subjects attempted to preserve their 90° ankle angle by maintaining a constant level of isometric muscular activity against the knee pad to support the load, following a perturbation to the mass. The resultant stiffness of these structures involved is estimated from the weight of the load and the timing or frequency from the dampened force oscillation pattern following the perturbation. Evidently, this technique is unable to identify relative contributions of other structures speculated to contribute to the movement (e.g. the hip flexors). Then again, isolating specific lower-body muscles in a challenge also faced by the dynamometry plantarflexion task discussed above. Furthermore, technical issues regarding delivery of the perturbation and the levels of attention focus from the participants during its delivery were also reported as potential inconsistencies of this method [89]. Consequently, direct imaging of structural elements under controlled conditions is viewed as a more specific method to investigate the role of tendon stiffness.

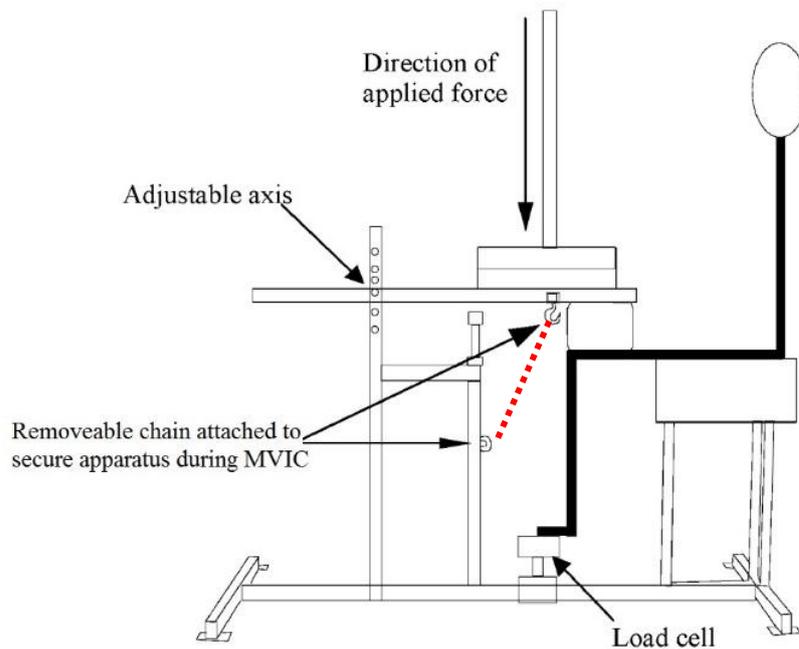


Figure 2-3. Schematic of instrumented seated calf raise machine used to determine unilateral maximal isometric force and lower-limb stiffness from the free oscillation model.

Source: Spurrs [16]

2.4.3. Methodological considerations in measuring component stiffness

Dynamometry measures the moment about one axis of the ankle. However, movement in a contraction occurs in 3 planes, particularly during a maximal effort. Moment values about the other two axes are typically assumed negligible [82, 90]. Nonetheless, due to the difficulty preventing this excess movement, even when using external fixations in an MVIC, its existence should be considered a limitation [15]. Attempts to account for the inevitable misalignment between the rotational axis of the dynamometer and ankle joint, several additional calculations have been proposed. Arampatzis et al. [91] described a method of measuring tendon displacement during a separate passive isokinetic trial (ankle moved independently of the participant). The corresponding change in tendon length from a known degree of ankle rotation can be determined. Subsequently, the required proportion of this elongation is then subtracted from a measured elongation during a MVIC, dependant on the ankle rotation observed in the MVIC [91]. The resultant joint moments can also be calculated using inverse dynamics from measures obtained in 3D analysis [91]. Magnusson et al. [92] also highlighted the significance of such corrections, reporting a 3° movement in ankle angle (when assumed to be zero) led to an additional tendon displacement of 2.6 mm without any change in the mechanical force exerted by

the tissue. Kongsgaard et al. [82] used an electronic goniometer positioned over the medial aspect of the ankle joint to identify joint moments and attempted to make necessary adjustments to length change. Kongsgaard et al. also reported that a correction for the observed ankle joint movement during MVICs ($0.6\text{--}1.4^\circ$) had a negligible effect on subsequent stiffness calculations. Future research should consider these extraneous movements while acknowledging that supplementary corrections to joint rotation add additional steps whereby measurement error can occur and be amplified.

A further analysis issue that varied between studies was the selection of the point on the force-length curve where mechanical properties represented. The slope from 50 to 100% [15, 81] or from 80 to 100% [82] of the MVIC, are examples of expressions of stiffness. The selection of these expressions is based on a previously hypothesised linear relationship between force and length in this upper region of the curve [92]. Considering submaximal running does not elicit maximal PF efforts, clarification surrounding the optimal range for analysis is still required. Tissues comprising an MTU are known to deform both longitudinally and transversely during loading [93], which cannot be accounted for with two-dimensional (2D) ultrasound. Although 3D ultrasound can allow for more accurate measures, investigations involving this expensive equipment have only recently commenced [94].

Finally, assumptions regarding bilateral stiffness by measuring only one MTU should be made with caution. Bohm, Mersmann [83] reported an asymmetry index range of 3-31% in measured AT stiffness in recreationally active young men. Despite running gait appearing symmetrical, different loading profiles of dominant and non-dominant legs need to be considered. Across research methodologies of stiffness, assessments are often standardised to one leg for an entire sample [95, 96], despite leg preference having an effect on $k_{\text{joint/ankle}}$ during hopping [97] and k_{leg} during running [29]. Studies which aim to quantify changes in tendon or global stiffness, as a result of an intervention or as an injury screening tool, need to consider bilateral measurements of stiffness.

2.5. Validity of lower-body stiffness measurements

Across the various global models discussed in part 2.3 of this review, it is generally agreed that $k_{\text{leg/dynamic}}$ is the current ‘gold standard’ for stiffness measurement in human running [2]. In a study of competitive male triathletes (N=12), Lorimer [48] reported this model

held excellent correlations with all levels of stiffness ($ICC = 0.97$), including with the ‘true’ method for estimating ΔL when calculating k_{leg} according to the method proposed by Colman et al. [2]. Additionally, multiple authors [2, 5, 21] now regarded $k_{leg/time}$ as a valid surrogate measure when 3D motion capture is not available or is not practical.

Comparability between hopping and running measures could potentially be limited for two possible reasons. First, the kinematics of vertical hopping usually elicit a forefoot landing, while in running there are varying foot strike patterns with degrees of rearfoot to forefoot landings [48, 98]. This effect influences the amount each joint stores and produces energy, thus affecting the association between hopping and running stiffness assessments [58]. A second factor emerges from the assessment of the SMM at near maximal running speeds. Clark and Weyand [39] observed deviations from the SMM waveform in an elite group of sprinters at maximal velocities ($10.36 \pm 0.27 \text{ m}\cdot\text{s}^{-1}$). This deviation was proposed to be from enhanced GRFs early in initial foot touchdown [99]. However, such observations may not extend to the measurement of k_{leg} during steady-state running or even maximal running by distance athletes who do not dedicate substantial training time to sprinting mechanics.

Hopping and isolated MVICs are not typically performed as training modalities for many athletes, bringing into question the validity of these assessments to reflect capabilities in dynamic movements such as running. There is a paucity of research that explores the relationships between the component and global stiffness assessments and to date, the findings were mixed. McLachan et al. [40] found no significant relationship between k_{vert} from bilateral hopping and stiffness from the vertical oscillation method in recreationally active females. When comparing genders, Hobara and colleagues [100] demonstrated that females showed a significant positive linear relationship ($P < 0.01$) between k_{MTU} during passive PF and k_{vert} during hopping at 2.5 ($r = 0.77$) and 3.0 Hz ($r = 0.83$) respectively. However, no significant relationship existed in males. The participants in this study were sedentary or only mildly active. Thus, its applicability to athletic performance is problematic. Using a different methodology to Hobara et al. [100], Rabita, Couturier and Lambertz [101] assessed well-trained gymnasts, long-distance runners and a control group to explore relationships between k_{vert} and k_{MTU} . Their study reported a non-significant relationship between measurements of intrinsic mechanical properties of the ankle and k_{vert} during bilateral hopping within any groups, or when the sample was pooled ($N=27$). Accordingly, the authors highlighted the importance of other factors aside

from the elastic components that determined jump task performance. Neuromuscular control and rate of force development were identified as potentially important factors to stiffness differences between groups [101].

2.6. Reliability of common methods of lower-body stiffness

The reliability of different stiffness measures should inform decisions on task or assessment selection, appropriate for the question of interest. Joseph and colleagues [45] examined vertical, leg, knee, and ankle stiffness in 20 active men and reported good and moderate reliability for k_{vert} and k_{leg} respectively ($\text{ICC} \geq 0.80$ and $\text{CV} \leq 10\%$), but poor reliability for knee and ankle stiffness during overground running. Additionally, both uni- and bilateral hopping (2.2 Hz) showed good reliability for k_{vert} but not for ankle or knee stiffness. Notably, it can be assumed that overground running used in this study, influenced inter-step variability, due to various speeds performed across their short 10 m running task. Joseph et al. reported the target was $3.89 \text{ m}\cdot\text{s}^{-1}$, with the true observed speed across trials being $3.35 \pm 0.12 \text{ m}\cdot\text{s}^{-1}$ demonstrating inter-individual and intra-individual variation.

Lorimer [48] performed a reliability study across all forms of global stiffness in 12 competitive triathletes. Her research found that over two testing occasions, inter-day reliability was ‘good’ for $k_{\text{leg/dynamic}}$ during running, with the difference between the means $< 5\%$ and effect size < 0.6 providing this interpretation. Assessing stiffness of the three individual joints in her sample exhibited weaker correlations with the expression of $k_{\text{leg/dynamic}}$ ($r = 0.20, 0.76, 0.17$) for ankle, knee and hip respectively, supporting the theory of high variability in the running task at the joint level. Interestingly, when Lorimer combined all three joints, or two out of the three (hip+knee or hip+ankle), variability was reduced to an acceptable level, and overall reliability was improved. An abbreviated summary of the reliability of global stiffness measures from this research [48] is provided in *Table 2-3*.

Due to the variety of techniques for the assessment of component stiffness and a lack of accurate replication of such techniques, it is difficult to state conclusively the reliability of this stiffness measurement. To utilize these methods for chronic tendon adaptation, future research should determine and interpret the reliability in the context of the signal to noise ratio of the measure as well as the smallest worthwhile change for influences on

sporting performance. Spurrs et al. [16] performed an intervention study with trained runners using a free oscillation method (part 2.4.2), yet did not report the inter-day reliability of their method, instead citing a study with similar technique [87] but with different apparatus. This original oscillation method paper by Walshe et al. reported good reliability ($r = 0.94$, CV 8%), and Spurrs and colleagues reported increased stiffness above the coefficient of variation (14.9% and 10.9% for the left and right leg respectively). McLachlan et al. [40] performed a further investigation into free oscillation, using an identical setup to Spurrs et al. [16], to assess its reliability across three sessions. McLachlan et al. reported levels of reliability between sessions two and three were acceptable (ICC = 0.93, ES = -0.33) but not between the initial and second assessments (ICC = 0.43, ES = -1.29). Thus, it was suggested that familiarization with the protocol is recommended [40]. Joseph et al. [81] presented a technical report on reliability of *in vivo* stiffness assessed by an ultrasound and isokinetic dynamometry set up (as described in part 2.4.1). In their sample of five males and five females, excellent reliability was reported for both the MVIC moment (ICC = 0.99, 0.95) and maximal aponeurosis elongation (ICC = 0.99, 0.93) for the intra- and intersession measures respectively.

Table 2-3. Reliability of commonly used models for assessing global stiffness as assessed in well-trained male triathletes (N=12)
Source: Lorimer [48]: reproduced with permission

Classification	Equation# (reference)	Model Name	Task	Reliability		Variability		Overall Reliability
				MDiff%(CL)	ES	CV% (CL)	ICC (CL)	
Vertical stiffness	#1 (McMahon and Cheng [1])	$k_{\text{vert/dynamic}}$ Dynamic Vertical stiffness	HP: 10 submaximal single leg @ 2.2 Hz - instructed to keep leg straight as possible	2.8 (-1.9 – 7.7)	-0.18 (-0.20 - -0.16)	5.8 (4.2 – 9.8)	0.91 (0.73 – 0.97)	Good*
	#2 (Morin et al. [21])	$k_{\text{vert/time}}$ Temporal estimated Vertical Stiffness	HP: 10 submaximal single leg @ 2.2 Hz - instructed to keep leg straight as possible	1.1 (-1.6 – 3.9)	-0.14 (-0.15 - -0.13)	5.8 (4.2 – 9.8)	0.85 (0.62 – 0.95)	Good*
Leg stiffness	#3 (McMahon and Cheng [1])	$k_{\text{leg/dynamic}}$ Dynamic Leg stiffness	TMR: 3.3 m·s ⁻¹ mean of 10 steps	0.2 (-2.5 - 3.0)	0.02 (0.00 – 0.03)	3.6 (2.7 – 3.8)	0.97 (0.90 – 0.99)	Good
	#4 (Morin et al. [21])	$k_{\text{leg/time}}$ Temporal estimated Leg Stiffness	TMR: 3.3 m·s ⁻¹ mean of 10 steps	0.3 (-1.8 – 2.4)	-0.00 (-0.01 – 0.01)	2.8 (2.0 – 4.4)	0.98 (0.93 – 0.99)	Good
Joint stiffness	#5 (Farley et al. [44])	$k_{\text{sumjoints}}$ Dynamic Joint Stiffness	HP: 10 submaximal single leg @ 2.2 Hz - instructed to keep leg straight as possible TMR: 3.3 m·s ⁻¹ mean of 10 steps	-5.4 (-10.5 – 0.0)	0.36 (0.32 – 0.40)	7.9 (5.8 – 12.5)	0.86 (0.65 – 0.95)	Moderate*

= Stiffness equations as presented in full in *Table 2-1*,

HP = hopping, TMR = treadmill running, ES = effect size, CL = 90% confidence limits, * at least one reliability parameter was unclear (confidence interval spanned more than one criteria). Average reliability was determined to be ‘good’ if the percent difference between means (MDiff%) was <5% and the effect size (ES) was *trivial* (0 – 0.2) or *small* (0.2 - 0.6) [102]. If one of these criteria were not met, then measurement reliability was interpreted as ‘average’. ‘Poor’ reliability resulted in neither criteria being met [103]. Measurement variability was assessed from typical error, reported as the coefficient of variation percentage (CV%) [103, 104] and intra-class correlation coefficient (ICC) [103]. Criteria for ‘small’ measurement variability were CV <10% [103] and ICC >0.70 [103, 105]. If CV was >10% or ICC <0.70 then variability in the measurement was deemed ‘moderate’. ‘Large’ measurement variability was reported if neither criteria for ‘small’ was met.

Overall reliability: was deemed ‘Good’ reliability if all four criteria to be met (ES, MDiff%, CV and ICC) ‘Moderate’ reliability resulted from one criteria outside the limits, while if two or more criteria were outside the limits, ‘poor’ overall reliability was recorded [45].

2.7. Practical Recommendations and Conclusion

All stiffness measures should be considered as a system in relation to each other, rather than in isolation, to gain a wider appreciation of their role in human performance. It can prove difficult to compare studies following the differences discussed in this review regarding the methodologies and calculations used. Notable disparities throughout the stiffness assessment literature include participants (gender, age, training status), 3D motion analysis methods and processing, ultrasound probe placements, running surface, running velocity and/or hopping frequency. From a practical standpoint, to enhance knowledge in this area, future studies should carefully consider the selection of each stiffness variable and its measurement technique, to ensure valid comparisons can be made. Additionally, familiarising participants with unaccustomed tasks to reduce the learning effect, may enhance measurement validity and reliability [40, 45]. The effect of pre-testing warm-up protocols also warrants greater exploration in the interest of increasing reliability of tendon stiffness. Finally, it appears longitudinal research is yet to be conducted utilizing k_{joint} to measure changes in lower-body stiffness across each joint during as a result of well-structured training interventions. Such studies would aid guidelines for levels of "optimal" stiffness that are specific for a sporting task at the joint level. Until such work is undertaken, suggesting the ideal level of lower limb stiffness an individual should possess during running, remains to be determined.

CHAPTER 3

LITERATURE REVIEW TWO

A review on lower-body stiffness as an indicator of distance running performance

3.1. Introduction

Distance running performance is complex and multifaceted, with determinants of success widely studied and reviewed [106]. As the knowledge underlying key athlete capacities is progressing, researchers are giving particular attention to understanding and enhancing mechanical movement efficiency, with lower-body stiffness measurement proposed to offer such insights.

From a classical physiology perspective, endurance abilities are commonly quantified using key aerobic capacities including peak oxygen uptake ($\dot{V}O_{2\text{peak}}$), the coinciding pace of $\dot{V}O_{2\text{peak}}$ ($v\text{-}\dot{V}O_{2\text{peak}}$) and various expressions of submaximal energy efficiency, commonly known as running economy (RE). Across highly-trained and elite distance runners, RE can surpass aerobic capacity ($\dot{V}O_{2\text{peak}}$) in predicting performance, varying by as much as 30% in athletes displaying similar levels of $\dot{V}O_{2\text{peak}}$ [107]. Within the middle-distance (MD) events of Athletics, runners must be capable of sustaining high velocities over the course of a race, often running at supra-maximal aerobic speeds (i.e. above $v\dot{V}O_{2\text{peak}}$) [6], with mean race velocities during competition reported as $7.65 \text{ m}\cdot\text{s}^{-1}$ in international male MD runners [5]. This highlights the need for a high $v\dot{V}O_{2\text{peak}}$, coupled with an even greater maximal anaerobic sprinting speed (v_{max}). It could be argued that this v_{max} is of great importance in determining medal winning ability, as analysis of MD events has demonstrated the significance of the final ‘kick’ in determining championship outcomes [8].

The examination of running through a mechanical lens, reveals that the musculoskeletal system responds accordingly to the impact forces of landing through the compression of a runners legs. As presented in Chapter 2, stiffness is the relationship between a force applied and the resulting compression of an object or body [18]. Throughout each gait cycle, the impact of the foot strikes are based on displacement or lowering of the runner’s CoM, achieved through the compression of their ‘leg-springs,’ originally termed the ‘bouncing gait’ [1, 4]. This early model described the leg during stance as a massless

mechanical spring, loaded by the body. At each ground contact, mechanical energy is stored in the elastic components of the leg as they lengthen eccentrically and ground reaction forces (GRFs) are absorbed, allowing a portion of this stored elastic energy to be utilized in the subsequent concentric propulsive phase [2, 21]. The action of muscles and tendons storing and returning elastic energy is represented in the stretch-shortening cycle (SSC) [108]. The SSC contribution is one of the key biomechanical mechanisms in contributing to RE and v_{\max} [16, 29, 109, 110]. Optimising the ‘free’ energy stored within the elastic components of the limbs, permits greater power in the concentric contraction, without additional increase in metabolic energy requirement [12, 111], hence energy ‘free’ from metabolic cost. As such, a ‘stiffer’ leg-spring theoretically is advantageous to efficient distance running performance [16, 29]. Effective return of energy from a stiffer spring is valued by sprinters and distance runners alike. Cavagna [110] examined stride-to-stride GRFs in sprinters and deduced that maximal power is sustained by the mechanical energy stored in the series elastic elements during ground contact, and released immediately in the drive phase, affording the athlete a greater v_{\max} .

In the current literature, there is a lack of consensus between contributions of certain biomechanical, physiological and anthropological elements that characterize the fast yet equally efficient MD runner [112, 113]. Lower-body stiffness is emerging as a commonly cited biomechanical influence to running performance, encompass SSC efficiency and furthers understanding by quantifying a runner’s ability to absorb and utilize energy throughout gait [16, 29, 37]. Therefore, the aim of this review is to examine the multiple methods of assessing stiffness used within the literature and the importance of its assessment in qualifying middle and long distance running performance.

3.1.1. Search Strategy

The PubMed, CINAHL, Web of Science and SPORTDiscus databases, to July 2015, were searched for terms linked with the Boolean operators (‘AND’, ‘OR’, ‘NOT’): ‘stiffness’, ‘run*’, ‘performance’, ‘economy’, ‘ultrasound’. Papers were selected based on the title, then abstract and finally text. Papers were included if they assessed trained middle- and/or long-distance runners, specifically included a measurement of stiffness or its related mechanical components, in addition to one or more direct or indirect performance measures from: RE, $\dot{V}O_{2\text{peak}}$, v_{\max} and simulated (i.e time trial) or actual race performance. However, further supporting research, such as training intervention studies involving other athletes (e.g. sprinters, triathletes) have been highlighted where deemed

appropriate, but have not been included in a tabulated comparison of studies. Papers were excluded if they focused on the following topics: prepubertal children, effect of footwear, surgery, treatment, diagnosis, and tendinopathy associated with injury or disease. Case reports, editorials, letters to the editor and animal studies were excluded. The review of original research contributing to the understanding of the links between stiffness and performance, will be discussing in the following sections, and is divided into three levels of evidence; descriptive, cross-sectional, and longitudinal.

3.2. Stiffness classification

Measurements of stiffness have differed substantially throughout the literature over the past four decades, both in the tasks performed, equipment employed and calculations made, each depending upon the specific questions and the laboratory resources available [18]. This information, as presented in detail in *Chapter 2*, is briefly summarised in *Table 3-1* below. The simplified SMM (global stiffness) exists to provide a current best estimate, for what is in reality a complex and intricate system of many anatomical structures and neuromuscular influences, making a true assessment highly impractical [18, 114]. Despite the existence of ‘gold-standard’ assessments for global stiffness (3D kinetic and kinematics) [2, 48], these do not directly quantify the contribution of all structural components (tendons, ligaments, muscles, cartilage, and bone). As such, isolation of lower body components to examining their material and mechanical properties has been employed [81, 115], despite several methodological challenge for researchers (see *Chapter 2*, 2.4.3).

Table 3-1. Summary of categories and definitions of lower-body stiffness

	Appropriate Term	Definition
<i>Component stiffness</i>	Muscle-Tendon Unit Stiffness (k_{MTU})	The displacement of the muscle relative to the tendon, of which segments can be imaged (<i>in-vivo</i>) alongside recording of the contributing forces of both the agonists and antagonists
	Joint Stiffness (k_{joint})	The resistance to rotation of a joint calculated via modelling the joints as torsional springs
<i>Global Limb Stiffness</i>	Leg Stiffness (k_{leg})	The resistance to change in leg length occurring from initial ground contact to the bottom of stance; a measure of individual leg springs
	Vertical Stiffness (k_{vert})	The measure of how the human body, as a whole, responds to the ground reaction force applied to the system where there is no horizontal displacement (i.e. hopping or jumping)

Adapted from *Chapter 2* and previous literature [114] [26]

3.3. Descriptions of stiffness during running

Several studies have described comprehensive profiles of well-trained athletes reporting a measure of stiffness [36, 116], providing normative values for certain groups of athletes (for a given methodology). These normative values can be used to track stiffness changes resulting from modified training prescription [117], or acutely across a run to exhaustion [118, 119]. Arampatzis et al. [36] published a report on the differences between Achilles tendon stiffness (k_{AT}) in sprinters and endurance runners. Sixty-six males, of which 28 were sprinters and 28 endurance runners, were compared with each other and to a reference group of ten adults not active in sports. A method of performing a plantarflexion MVIC in a dynamometer was employed. Sprinters had a significantly higher ($P < 0.05$) normalised k_{AT} ($1.97 \pm 0.39 \text{ Nm}\cdot\text{kg}^{-1}$) than the endurance runners ($1.63 \pm 0.46 \text{ Nm}\cdot\text{kg}^{-1}$) and the non-active adults ($1.41 \pm 0.29 \text{ Nm}\cdot\text{kg}^{-1}$). Although not correlated to the runner's v_{\max} , this stiffness data gives an indication that greater stiffness in the AT may be beneficial, if not required, for higher velocity movements compared to those who perform mainly endurance running. One could assume sprinters are exposed to greater lower-body forces; as a function of higher velocity running, and a greater portion of time performing resistance training, compared to endurance runners. This may explain their larger stiffness values [120] although training regimes were not reported. Similarly, Hobara et al. [116] examined k_{vert} between endurance (N=7) and power-trained athletes (N=7) during 15 consecutive bilateral hops (1.5 and 3 Hz). The difference in k_{vert} between the two groups was attributed to the power athletes increased k_{joint} at their knee and ankle. This study also included EMG of six lower-body muscles, observing greater activity in the endurance cohort when differences between the groups were significant. These authors concluded that differences in k_{joint} between groups were subject to factors other than the neuronal input or touchdown joint angle, and potentially modulated by the intrinsic properties of the tendon itself. While comparative studies are unable to surmise a specific link between stiffness and training modality, such literature supports the role of training stimuli in stiffness adaptation nonetheless. These studies aid in clarification of such mechanical requirements for successful performance, specific to the task or event.

Lower-body stiffness maintenance during running, may hold the potential to predict running performance. Hayes and Caplan [118] showed in a small sample (N=6) of sub-elite males MD runners, that failure to maintain k_{leg} while running at $v\cdot\dot{V}O_{2\text{peak}}$, lead to reduced performance. Specifically, changes in k_{leg} rather than k_{vert} were correlated to both the distance covered and the duration of a run to exhaustion performed at $v\cdot\dot{V}O_{2\text{peak}}$ ($r =$

0.87 and 0.83, respectively). Interestingly, these authors reported that changes in k_{leg} provided a better correlation to performance than their metabolic measures ($\dot{V}O_{2peak}$, lactate threshold, and RE), accounting for 75% of the variance in distance covered and 68% of the variance in time to exhaustion at $v\text{-}\dot{V}O_2$. The influence of different training regimes amongst the small sample was acknowledged, with four of the six athletes regularly performing resistance training. Furthermore, in their earlier study [121], Hayes and Caplan found during MD competition an increase in contact time was observed across each lap, indicating neuromuscular fatigue may increase contact times and presumably decrease stiffness.

Longitudinal descriptive studies can highlight the effect of training (or de-training) on stiffness. Kobu et al. [117] examined RE and MTU properties in 11 well-trained long-distance runners (mean 5000 m time 14:33, min:s) and six untrained controls (matched for body dimensions) across the track and road racing seasons (in that order). The tendon stiffness of both the knee extensors and plantar flexors were measured alongside other lower-body assessments of muscle strength, neuromuscular properties and jump performances. Stiffness was significantly ($P < 0.05$) lower in both knee extensors (-14.4%) and plantar flexors (-16.6%) at the road racing time point (higher running volume), compared with the previous testing during the runners track preparation (lower volume, greater intensity). Improvement in RE at three submaximal velocities (14, 16 and 18 $\text{km}\cdot\text{hr}^{-1}$) was also observed at the later time point. Enhanced RE during the road racing season, may be due to the observed increase in the compliance of the tendon structures at both sites, without alterations in neuromuscular characteristics [117]. However, it is speculated that potential cardiopulmonary adaptations may have resulted in the RE improvements (yet $\dot{V}O_{2peak}$ was not reported), owing to a significant increase in monthly running volume (12.7%, ES = 1.31) between track (718 \pm 80 km) compared with road season (832 \pm 95km).

3.4. Relationships between stiffness and running

Lower-body stiffness, as both a global and component quantity, has been assessed in the literature in trained runners (see *Table 3-1*). Considering global assessments of stiffness in regional level MD runners, Dalleau et al. [29] reported a significant relationship between RE at a submaximal speed (90% of $v\text{-}\dot{V}O_{2peak}$) and k_{leg} ($r = 0.80$, $P < 0.05$). The k_{leg} of both limbs was calculated from measurements made via a kinematic arm, with the

reference end of the arm fixed to the ceiling, while the moving end was attached to the back of the participant (CoM) running on the treadmill. The runners displacement (vertical and horizontal) were derived from the kinematic arm and stiffness calculated via measurement of positive and negative work. This novel research paved the way for subsequent measures of stiffness via the SMM in endurance runners.

Barnes, McGuigan, and Kilding [31] combined 63 well-trained male and female runners, who competed in distances from 800 m to 10,000 m (see *Table 3-2*). Comparing genders, they found that k_{vert} , during a continuous bilateral hopping task (5 maximal efforts) had a *large* ($r = -0.57$) and *very large* ($r = -0.76$) correlation to RE in males and females respectively. An indication of a non-modifiable mechanical influence on RE was also published, namely the moment arm (MA) of the ankle joint. Lacour and Muriel [123] described a shorter MA results in a greater force in the lengthening AT during landing, thus increasing the amount of potential stored energy in the tendon, reutilized during push off. While Barnes et al. [31] observed a *very large* ($r = 0.81$) relationship between MA and RE, they stated that a single lower-body measure cannot exclusively explain differences in RE between individuals or genders.

Component measures of stiffness were examined by Fletcher, Pfister and MacIntosh [33] with a sample of 11 male and 18 female recreational road runners. Their results are synonymous with previous findings examining global stiffness [31], showing disparities in stiffness and RE relationships across genders. In the women, their relationship of k_{AT} (at 25–45% of MVIC) and RE was significant at all measured speeds ($r = 0.45$ $P < 0.05$), yet this was non-significant for men, possibly influenced by their smaller sample size. Combining genders, Fletcher et al. revealed k_{AT} showed a *moderate* correlation ($r = 0.40 - 0.44$) with RE expressed as a relative energy cost (E_C) ($\text{kcal} \cdot \text{kg}^{-1} \cdot \text{km}^{-1}$) at all relative submaximal speeds (75, 85 and 95% of lactate threshold). Including submaximal speeds offers further insight into the specific contribution of the aerobic energy system to runners with differing levels of k_{AT} and across genders. Additionally, Fletcher et al. cited their previous study [124] by speculating this tendon stiffness is finely tuned to minimise the shortening of the muscles in series with it, thereby reducing the metabolic requirements during submaximal running intensities.

Muscle strength and power are intrinsically linked to stiffness due to their influence on the SSC [19, 125]. Using the free oscillation method (see *Chapter 2, 2.4.2*), Dumke et al.

[30] made an indirect assessment of the musculo-articular stiffness of the ankle joint using the free-oscillation technique in a sample of 12 highly-trained male runners (mean 5 km time of 15:04, min:s). Stiffness was significantly correlated to RE at the fastest submaximal running speed (19.3 km·hr⁻¹, $r = -0.69$, $P = 0.01$), yet not at any slower speeds (11.25 – 17.7 km·hr⁻¹), in agreement with Dalleau et al. [29] when measuring k_{leg} . Dumke et al. assessed lower-body power (countermovement and squat jumps) and strength (isometric squat) and showed non-significant correlations between stiffness quantified via the seated calf-raise task, while significant associations amongst countermovement jump power and RE were observed at three submaximal speeds of 12.9 ($r = 0.66$), 14.5 ($r = 0.70$), and 16.1 km·hr⁻¹ ($r = 0.58$). Unfortunately, k_{vert} from the jumping assessments was not calculated. A SMM calculation in this instance would have provided insight into the link between the static oscillation task and the influence it has on dynamic movements.

Components along the kinetic chain act differently in running than in static tasks, thus, accurate predictions of performance from measures MTUs in isolation may be limited. Studies examining multiple MTUs in the same population can however provide improved insight into how elastic energy is stored through multiple sites. Arampatzis and his colleagues [32] grouped 30 runners based on RE values and reported the sub-group with superior efficiency (N=10) displayed stiffer MTUs in the TS and more compliant MTUs at the quadriceps femoris (QF), compared to the groups with moderate and poor RE. These differences were reported despite non-significant differences in kinematic parameters (joint angles of the hip, knee and ankle) during submaximal running. As such, it can be speculated that isolated 2D kinematic measures may not accurately define differences in measured metabolic efficiencies between runners. Additionally, no significant difference in this samples passive morphological properties across both the TS and QF components (fascicle length, angle of pennation, and tendon thickness) were observed. A potential mechanism behind this observation was proposed in studies investigating *in-vivo* fascicle behaviour during submaximal running [77-79]. Examining Kenyan and Japanese national-level runners, Sano et al. [77] found during the first half of stance, shortening of the medial gastrocnemius fascicles occur while the series elastic components and the MTU lengthen as a whole. This study noted that longer ATs of the Kenyans may be considered advantageous in preventing the MTU stiffness from declining during the repeated cyclic movements, thus enhancing their performance. These authors also revealed no differences in contact time between the two ethnicities at slower

speeds (up to $13 \text{ km}\cdot\text{hr}^{-1}$) however, the examination of faster speeds were not performed. Therefore, the examination of MTU components does further the insight into the differences in metabolic cost between runners, with supporting evidence that a stiffer MTU undergoing less strain in the lower-body may be superior for more efficient distance running [32].

The above component stiffness study is contrary to results in a recent publication by Kubo and his colleagues [126], notably in a much larger sample of male long-distance runners ($N=64$). Defined by best their 5000 m performance (see *Table 3-2*), the faster athletes in this study showed stiffer tendon structures in the knee extensors and more compliant structures in TS compared to the slower 5000 m runners. Comparing the results above of Arampatzis and his colleagues [32] to these of Kubo et al. [126], the definition of running performance, i.e; RE [32] versus race time [126], may explain the conflict in the direction of the relationship with stiffness. Methodological differences in the MTU assessments between these two studies, also highlights a lack of consensus on stiffness and compliance, especially when assessing components in isolation. Therefore, differences in measures of the TS or QF, cannot confidently explain the differences in RE within trained runners. The argument therefore, would be to perform component assessments alongside dynamic measures of global stiffness to further the understanding of how isolated components affect whole kinetic chain movements.

Table 3-2. Cross-sectional studies of lower-body stiffness in trained distance runners

Study (y) [reference]	Sample (N)	Training volume, Aerobic capacity ($\dot{V}O_{2peak}$ in $ml \cdot kg^{-1} \cdot min^{-1}$)	Measurement(s) of Stiffness (k)	Measurement(s) of Running Performance	Relationships between Measures
Arampatzis et al. (2006) [32]	28 M; LDT [split into 3 groups based on RE using a cluster analysis]	TV: 40-120km/wk AC: no measured	<p>» Stiffness <i>strain</i> ($kN \cdot strain^{-1}$) of tendon structures in knee (QF) extensors & plantar flexors (TS) during MVIC using DYNO</p> <ul style="list-style-type: none"> - High RE: 34.4 ± 6.6 - Med RE: 23.7 ± 9.6 - Low RE: 20.6 ± 6.6 <p>» SS differences* in between High RE group and the other two groups</p>	<p>» RE at 3, 3.5, 4 $m \cdot s^{-1}$ ($mL \cdot kg^{-1} \cdot min^{-1}$) [High, Med, Low RE group]</p> <p>3 $m \cdot s^{-1}$: [37.4 ± 1.9, 39.9 ± 1.8, 44.4 ± 1.9]</p> <p>3.5 $m \cdot s^{-1}$: [43.6 ± 1.3, 45.7 ± 1.6, 49.6 ± 2.5]</p> <p>4.0 $m \cdot s^{-1}$: [48.9 ± 2.0, 52.0 ± 1.5, 58.9 ± 1.5]</p>	<ul style="list-style-type: none"> ► Group with high RE showed SS* greater compliances of QF stiffness and higher TS stiffness compared to the moderate and low RE groups ► Neither the kinematic parameters nor tendon morphological properties showed SS* differences between groups
Barnes et al. (2014) [31]	39 M 24 F MDT + LDT	TV: M = 97.2 ± 21.0 $km \cdot wk^{-1}$, F = 74.2 ± 12.7 $km \cdot wk^{-1}$, AC: M = 68.7 ± 4.8 F = 59.9 ± 3.5	<p>» k_{vert} ($kN \cdot m^{-1}$) from 5-H test</p> <p>M = 9.4 ± 2.2</p> <p>F = 13.3 ± 2.7</p>	<p>» RE: at 3.89 $m \cdot s^{-1}$ no means reported</p> <p>» $v \cdot \dot{V}O_{2peak}$ from treadmill M = 21.1 ± 1.6 $km \cdot hr^{-1}$ F = 19.4 ± 1.2 $km \cdot hr^{-1}$</p>	<ul style="list-style-type: none"> ► k_{vert} and RE negative correlation M: $r = -0.57$; F: $r = -0.76$ Combined genders: $r = -0.80$ ► k_{vert} as showed a <i>large</i> negative relationship to moment arm length ($r = 0.82$)
Dalleau et al. (1998) [29]	8 M; MDT	TV: not reported AC: 65.6 ± 4.6	<p>» k_{leg} ($kN \cdot m^{-1}$) during treadmill running at 90% of individual $v \cdot \dot{V}O_{2peak}$: 5.10 ± 0.33 $m \cdot s^{-1}$</p> <ul style="list-style-type: none"> - ‘Propulsive leg’ = 18.8 ± 5.6 - ‘Stick leg’ = 22.3 ± 9.9 	<p>» RE: 175 ± 6 $mL \cdot kg^{-1} \cdot km^{-1}$</p>	<ul style="list-style-type: none"> ► RE and k_{leg} showed strong relationship ($r = 0.80^*$) ► Comparison of the legs revealed a stiffness asymmetry (labelled as stick & propulsive) but was statistically NS
Dumke et al. (2010) [30]	12 M; LDT	TV: 96.8 ± 23.6 $km \cdot wk$ AC: 68.3 ± 4.3	<p>» k_{MTU} ($kN \cdot m^{-1}$) determined from oscillation pattern from equation for motion of dampened spring [87]</p> <ul style="list-style-type: none"> - Muscle k of TS: 0.762 ± 0.275 - Tendon k of TS: 0.288 ± 0.098 	<p>» Race time 5000 m PB [min:sec]: $15:04 \pm 0:53$</p> <p>» RE : not reported</p>	<ul style="list-style-type: none"> ► Correlation ($r = -0.69^{**}$) between RE at the 19.32km/hr stage and muscle k ► NS relationship between k and other submaximal stages ► NS relationship between either CMJ or SJ variables with stiffness.

Table 3-2. Cont. Cross-sectional studies of lower-body stiffness in trained distance runners.

Study (y) [reference]	Sample (N)	Training volume, Aerobic capacity ($\dot{V}O_{2peak}$ in $ml \cdot kg^{-1} \cdot min^{-1}$)	Measurement(s) of Stiffness (k)	Measurement(s) of Running Performance	Relationships between Measures
Fletcher et al. (2013) [33]	11 M; 18 F; REC runners	TV: min 5 runs·wk (volume not measured) AC: M = 55.5 ± 0.8 F = 49.8 ± 0.6	» k_{AT} ($N \cdot mm^{-1}$) during MVIC in DYNO - Expressed as liner slope using 3 different force ranges as % of MVIC: [mean ±std error of the mean] - 25–45%: M = 164 ±8, F = 97 ±4 - 30–70%: M = 175 ±6, F = 108 ±5 - 50–100%: M = 191 ±5, F = 125 ±5 SS differences* between M & F at all ranges	» RE as E_C ($kcal \cdot kg^{-1} \cdot km^{-1}$) at 75, 85, 95% of the speed of lactate threshold 75%: M = 1.01 ±0.06, F = 1.05 ±0.10 85%: M = 1.04 ±0.07, F = 1.07 ±0.09 95%: M = 1.07 ±0.07, F = 1.09 ±0.10	► Differences in E_C not explained solely by differences in AT stiffness. ► M only: no correlation between k_{AT} and RE at any force ranges: ► F only: correlation between 25-45% MVIC k_{AT} and RE at all speeds $r^2 = 0.198^*$ ► M + F: correlation between k_{AT} and RE at all speeds ($r = 0.400-0.437^*$)
Kubo et al. (2010) [34]	15 M; LDT 21 M; untrained comparison group	TV: 150-200 km·wk. AC: not measured	» k_{MTU} ($N \cdot mm^{-1}$) expressed as liner slope above 50% MVIC performed in DYNO - knee extensors: RG: 70.5 ± 22.3 COM: 74.8 ± 22.9 - plantar flexors: RG: 31.1 ± 10.0 - COM: 34.9 ± 10.8	» Race time 5000 m PB [min:sec]: 14:32 ± 0:21 » RE: Not measured	► 5000 m PB was SS* correlated with <i>lower</i> k_{MTU} for both sites: plantar flexors $r = 0.77$ and knee extensors $r = 0.69$ ► NS difference in neural activation levels between trained and untrained groups
Kubo et al. (2015) [126]	64 M; LDT [split into fast / slow groups based on 5,000 m PB]	TV: min 6 runs·wk (volume not measured) AC: not measured	» k_{MTU} ($N \cdot mm^{-1}$) expressed as liner slope above 50% MVIC performed in DYNO - knee extensors: Faster group: 70.0 ±19.9 Slower group: 59.0 ±17.5 - plantar flexors: Faster group: 28.1 ±7.9 Slower group: 33.8 ±9.1 SS differences* between fast and slow	» Race time 5000 m PB [min:s] Faster group: 14:34 ± 0:10 Slower group: 15:14 ± 0:25 » RE: Not measured	► 5000m race PB was correlated to k_{MTU} in knee extensors: $r = -0.31^{**}$ plantar flexors $r = 0.41^{***}$ ► ultrasound method took into account relative contribution of physiological cross- sectional area medial gastrocnemius muscle for the calculation of k_{MTU}

M = male, F = female, REC = recreationally trained, MDT = middle distance trained (800-5000 m), LDT = long distance trained (5000 m +), TV = reported training volume at time of study, AC = aerobic capacity ($ml \cdot kg^{-1} \cdot min^{-1}$), PB = personal best race time, k_{MTU} = stiffness of a muscle-tendon unit, k_{AT} = Achilles tendon stiffness, TS = triceps surae, QF = quadriceps femoris, k_{leg} = leg stiffness, k_{vert} = vertical stiffness, MVIC = maximal voluntary isometric contraction, DYNO = isokinetic dynamometer, RE = running economy, E_C = total energy cost to run per unit of distance.

Data are mean ± SD unless indicated, SS = statistically significant at the following level: * $P \leq 0.05$, ** $P \leq 0.01$, or *** $P \leq 0.001$. NS = not statistically significant change

3.5. *Interventions to alter stiffness*

Researchers have demonstrated that non-running tasks, such as various forms of traditional [26], explosive [16] or isometric resistance training [17], may serve as an effective stimulus for improving running economy and distance running performance. Concomitant changes in stiffness are cited alongside these change in RE and performance in a number of studies discussed in the following sections.

3.5.1. *Acute effects*

Longitudinal lower-body stiffness research has determined that improved RE is linked to underlying neuromuscular properties thought to result in the better use of stored elastic energy [14]. Whether such neuromuscular influences can be favourably altered in an acute sense, was recently investigated by Barnes et al. [13]. Using a cross-over design where 11 well-trained male distance runners (see *Table 3-3*) served as their own control, the effects of a warm-up protocol with a weighted vest (20% body mass) on performance were assessed. Performing six 10 s ‘strides’ at a self-perceived 1500 m race pace in the weighted-vest condition, had a *very-large* effect of $v\text{-}\dot{V}O_{2\text{peak}}$ (2.9%; $\pm 0.8\%$), a *moderate* effect on k_{vert} (20.4%; $\pm 4.2\%$) and a *large* effect in RE at $14\text{km}\cdot\text{hr}^{-1}$ (6.0%; $\pm 1.6\%$). Analysis of the relationships between the change scores indicated altered k_{vert} could explain all the improvements in $v\text{-}\dot{V}O_{2\text{peak}}$ and RE. Despite a *small* negative effect on perceived race readiness scores, vest loading resulted in improved performance measures, presenting a potential tool to enhance endurance performance acutely [13]. The explanation for these changes were related to the post-activation potentiation (PAP) phenomenon, whereby the performance of skeletal muscle is influenced acutely by its contractile history [127]. From this work, it can be deduced that introducing an external load into chronic training, may increase global measures of stiffness and promote positive performance outcomes. The acute effects of different footwear may have inflicted the observed differences in stiffness in a recent study by Logan et al. [128]. It was reported that both spikes and racing flats caused vertical stiffness to increase compared to standard shoes, with spikes having a significantly greater effect on k_{vert} from increased GRFs, assessed via a synthetic track placed over a force platform. As such, Logan et al. [128] cautioned coaches to be mindful when transitioning footwear, especially after prolonged periods without spiked shoe running, where a sudden increase GRF and vertical stiffness could be potentially harmful.

3.5.2. Chronic training effects

Research involving controlled and counter-balanced training studies have offered support for the close association between lower-body stiffness and running performance. Resistance training may serve to enhance stiffness and ultimately performance, through several modes of training [129]. These can be divided into the following: 1) traditional or heavy strength training, 2) plyometric or explosive/ballistic training, and 3) isometric training.

3.5.2.1. Traditional or heavy strength training

Heavy or 'traditional' strength training (HST) appears linked to improvements in RE across several studies using well-trained populations [26, 130]. Few, however, have calculated and reported stiffness changes as a potential mediator for enhancement of metabolic efficiency. In a 14 week HST intervention, Millet et al. [26] implemented standardised lower-body free and machine weight exercises with highly-trained triathletes, two times per week while continuing their endurance training. Compared to the control group, HST resulted in significant increases in maximal squat and calf raise 1RM strength, hopping power, and notably RE and $\dot{V}O_{2\text{peak}}$. The k_{vert} from 10 s of continuous maximal hopping (2 Hz) was reported, yet did not show a significant change. The HST group did exhibit a moderate correlation for their change in RE with the change of hopping power ($r = -0.55$) following the intervention. With the number of exercises in this program, it is difficult to surmise the specific adaptations underpinning enhanced running performance, and whether the k_{vert} measure was sensitive enough to detect worthwhile changes following HST. Several years later, in an attempt to single out one HST exercise, Storen et al. [130] reported half-depth back squats alone, were sufficient to produce a significant 5% ($P < 0.05$) improvement in RE. This mixed gender study had a squat intervention group (N=8) perform reps at 4RM, three times per week and were compared with a control group (N=9) who maintained normal training. After eight weeks, the HST group showed a 26% improvement in rate of force development ($P < 0.01$), speculated to be partly responsible for the improvements in RE. As no direct measures of stiffness were calculated, in addition to these participants being recreational (mean 5000 m time 18:42; min:s), this work provides rationale for further examination into whether a simple and time efficient strength training program may improve global strength properties and if stiffness can quantify this alongside a change in distance running performance.

3.5.2.2. Plyometric and explosive training

Explosive-strength or plyometric training (PLY), comprising of ballistic jumping, hopping, or bounding, has been reported to promote specific neural adaptations suggested to improve RE [131]. Studying well-trained male runners, PLY interventions with matched control groups, have reported increased k_{MTU} [16] and k_{vert} [14] of the lower limbs, potentially liable for improved RE. In a brief six-week intervention, Spurrs, Murphy and Watsford [16] reported a PLY group improved in a 3-km time trial performance by 2.7%, ($P < 0.05$) with no significant change in a control group (*Table 3-3*). Saunders et al. [131] later reported similar changes in a higher sub-elite cohort of runners ($\dot{V}O_{2peak}$ 71.1 ± 6.0 ml·min⁻¹·kg⁻¹). These athletes enhanced their RE values by 4.1% following nine weeks of PLY. There was a non-significant but notable movement towards a greater average power during a 5-hop plyometric test (15%, $P = 0.11$) and a moderate increases in average jump height ($ES = 0.6$), following training. However, Saunders et al. did not model stiffness. Unlike Spurrs, Murphy and Watsford [16] who had elicited significant improvement in RE across slower submaximal velocities, the runners assessed by Saunders et al., only displayed improvements in RE at 18 km·h⁻¹ and not 14 or 16 km·h⁻¹. It is speculated that improvement in muscle power from PLY may assist in efficiency at higher speeds where a greater propulsive force per foot strike is required [16]. Furthermore, similarities between the angle of the foot at ground contact during faster velocity running and PLY based training exercises [132], would suggest a contribution of lower limb elastic recoil may be dependent on the degree of stretch across the tendon.

Barnes et al. [14] conducted a comprehensive combined gender study, incorporating two different strength training regimes during a cross country season (5 – 10,000 m events) in colligate runners. In addition to their typical endurance training, runners performed either pure HST or PLY combined with HST (PLY+H), with groups matched for volume and load. Race performances across the 13 week season were compared to athletes from other college teams whom the experimental athletes competed against, which served as a comparative group. Based on mechanistic inferences reported elsewhere [102], this study showed PLY+H negatively affected performance times for males by 0.8% ($90\% \pm 1.5\%$) with HST alone being possibly harmful ($0.1\%; \pm 1.3\%$). Conversely, in females; a likely beneficial effect of both PLY+H ($1.1\% \pm 1.3\%$) and HST ($1.4\% \pm 1.4\%$) on race times was shown compared with control runners. All athletes in the HST group showed a significantly greater increase in $v \cdot \dot{V}O_{2peak}$ (4.6% and 4.4% in men and women,

respectively), that the PLY+H group (1.0% and 2.2% in men and women, respectively). Notably, in agreement with previous studies [26, 130], HST also led to superior improvements in RE by 1.7% and 3.4% in men and women, respectively compared to PLY+H. All RE improvements occurred in the absence of any significant change in $\dot{V}O_{2peak}$, supporting the premise that changes were largely due to adaptations in neuromuscular characteristics rather than improved cardiorespiratory efficiency from endurance running. Unlike Spurrs et al. [16], who showed increased stiffness via PLY improved RE, Barnes et al. [14] stated that combining PLY and HST led to a mean reduction in k_{vert} ($-3.0 \pm 22.5\%$) in their male cohort, compared with HST only which increased lower-body stiffness ($+15.0 \pm 20.7\%$). Thus, training modality performed in this study had a moderate effect on stiffness. The suggestion that increased maximal strength underpins improved RE [26, 129], was also supported by Barnes et al. who elicited 20 to 40% improvements in runner's leg-press predicted 1RM, though improvements were 30 to 50% greater from HST. The group with the smallest improvement in maximal strength was males in the PLY+H group and furthermore was the only subset not to produce an associated increase in stiffness. This suggests HST is a superior stimulus for stiffness changes over time. It was acknowledged by Barnes et al., differences between the effects of training by gender, could partially be due to differences in training intensity during the season (i.e. training time run at $\geq 80\%$ of $\dot{V}O_{2peak}$), the required competition distances between men (8 to 10 km) and women (5 to 6 km), and the runners prior strength training experience.

3.5.2.3. Isometric Training

Several studies have examined components of the lower body to further the understanding of how these produce favourable stiffness and running performance adaptations [15, 17]. As the methods for tendon assessment commonly employ isometric contractions, the effect of isometric training (ISO) on modifying such components has also been explored. In recreational runners, Albracht and Arampatzis [15] revealed that an ISO protocol (four sessions per week, 14 weeks) resulted in significant 7% and 16% increases in ankle PF strength and k_{AT} respectively. These lower-limb changes were concurrent with a 4% reduction in RE ($P = 0.02$). This ISO stimulus was performed as five sets of 4 reps (3 s on; 3 s off) of ankle PF at 90% MVIC, adjusted week-to-week. In contrast, Fletcher et al. [17] found that in highly-trained runners, the effect of chronic ISO on k_{AT} and subsequently RE was not clear. Their prescription of longer contractions (4 x 20 s at 80% MVIC), did not result in a mean increase in k_{AT} ($0.9 \pm 25.8\%$) nor a mean improvement

in RE ($0.1 \pm 3.6\%$) in either the intervention or control runners. This could be owing to a shorter training duration (8 weeks) used by Fletcher et al. compared to the former 14-week study [15]. However, across the pooled participants, Fletcher et al. reported a moderate negative relationship ($r = 0.66, P = 0.02$) between the change in k_{AT} and change in RE, regardless of the assigned group (see *Table 3-3*). Furthermore, Albracht and Arampatzis [15] found the change in component stiffness following training in recreational men and women, did not coincide with any changes in spatiotemporal gait measures following ISO. Contact and flight time are incorporated in the modelling of k_{leg} [21], thus, global stiffness models may not detect changes following ISO as k_{leg} is influenced more by SSC capacities. This observation highlights differences between the two types of stiffness quantification and the countless mechanisms that influence efficient storage and utilization of elastic energy in running.

Table 3-3. Effects of training interventions on lower-body stiffness and running performance

Study (y) [reference]	Sample	Training volume, Aerobic capacity ($\dot{V}O_{2peak}$ in $ml \cdot kg^{-1} \cdot min^{-1}$)	Intervention Type	Stiffness Measurement	Δ Stiffness	Δ Running Performance	Additional Findings/Notes
Albracht & Arampatzis (2013) [15]	INT: 13 CON: 13 (Gender N/R) REC	INT: TV: 66 \pm 29 km·wk AC: N/R CON: TV: 62 \pm 31 km·wk AC: N/R	INT: 14 wks: ET + ISO 4/day/wk as ISO ankle plantarflexions 5 sets of 4: 3 sec on/off CON: Continued normal ET	$\gg k_{AT}$ ($N \cdot mm^{-1}$) expressed as linear slope above 50% MVIC performed in DYNO	INT: $\uparrow \sim 16\%$ CON: not measured	\gg RE: as E_c ($J \cdot m \cdot kg^{-1}$) INT: $\uparrow 4\%$ SS* CON: NSC	\blacktriangleright Running kinematics unchanged following ISO intervention \blacktriangleright Measured right leg only
Barnes et al. (2015) [35]	11 M MID + LTD	TV: 105.6 \pm 30.9 km·wk AC: 62.5 \pm 6.3	Cross-over design: \gg Acute INT: 6 x 10-sec strides @ perceived 1500 m race pace with loaded vest (~20% body mass) OR no external (vest) load [7-day washout]	$\gg k_{vert}$ from 5-H (bilateral, straight knee)	\gg Vest = moderate \uparrow = 20.4 \pm 4.2% [mean \pm 90% CI]	Vest INT: \gg RE: Large \uparrow of 6.0 \pm 1.6% $\gg \dot{V}O_{2peak}$: Very-large \uparrow of 2.9 \pm 0.8% [mean \pm 90% CI]	\blacktriangleright Vest had <i>small</i> negative effect on perceived race readiness score following strides, but with overall benefit to performance
Barnes et al. (2013) [14]	INT: 23 M & 19 F CON: Athletes from other teams MID & LTD	HRT group: TV: M = 91.9 \pm 12.1; F = 72.3 \pm 13.3 AC: M = 63.7 \pm 4.7; F = 52.3 \pm 3.3 PLY & HRT group: TV: M = 93.7 \pm 5.0; F = 73.6 \pm 13.8 km·wk AC: M = 63.8 \pm 4.6, F = 51.3 \pm 2.8	0- 5wks: normal ET INT: 5-13 wks of HRT <i>or</i> HRT+PLY 2/day/wk Mix upper/lower body exercises	$\gg k_{vert}$ from 5-H (bilateral, straight knee)	Male: HRT: \uparrow 15.0% HRT+PLY: \downarrow 3.0% Female: HRT: \uparrow 11.5% HRT+PLY: \uparrow 4.5%	\gg RE: HRT: M = \uparrow 2.1% F = \uparrow 3.4% HRT+PLY: M = \uparrow 0.2% F = \uparrow 1.0%	\blacktriangleright In some males, both interventions had negative effect on performance times \blacktriangleright <i>Small to moderate</i> reductions in body fat were observed despite no significant change in body weight
Fletcher, Esau, & MacIntosh (2010) [17]	INT: 6 M CON: 6 M ET	INT: TV: 108.5 \pm 41.9 km·wk AC: 67.3 \pm 3.1 CON TV: 107.9 \pm 40.8 km·wk AC: 67.6 \pm 6.4	INT: 4 x 20 s isometric contractions at 80% of maximum voluntary plantarflexion 8 wks: 3/day/wk CON: Continued normal ET	$\gg k_{AT}$ expressed as linear slope of 25-45, 30-70 & 50-100% of MVIC performed in DYNO	INT: NSC CON: NSC	INT: NS NSC CON: NSC	\blacktriangleright k_{AT} was greater when measured at a higher force range. \blacktriangleright SS relationship ($r = 0.66^*$) between the change in k_{AT} and change in RE, regardless of the assigned group when all force ranges of the MVIC and speeds for RE were combined

Table 3-3. Cont.

Effects of training interventions on lower-body stiffness and running performance.

Study (y) [reference]	Sample	Training volume, Aerobic capacity ($\dot{V}O_{2peak}$ in $ml \cdot kg^{-1} \cdot min^{-1}$)	Intervention Type	Measurement of Stiffness	Δ Stiffness	Δ Running Performance	Additional Findings/Notes
Millet et al. (2002) [26]	INT: 7 M CON: 8 M ET triathletes	INT: TV: 48 \pm 7 km·wk (running) AC: 69.7 \pm 3.6 CON: TV: 62 \pm 31 km·wk (running) AC: 67.6 \pm 6.4	INT: 14 wks: ET+ HRT As 2 x warm up sets, & 3-5 sets of 3-5 reps to failure, 2/day/wk CON: Continued normal ET	$\gg k_{vert}$ from hopping (2 legs) at 2 Hz for 10sec	INT & CON: NSC (SS* \downarrow in hopping contact times, & SS*** \uparrow hopping power)	$\gg v \cdot \dot{V}O_{2peak}$: INT: SS \uparrow ES = 0.57 CON: NSC \gg RE at 75% $v \cdot \dot{V}O_2$: INT: SS \uparrow ES = 1.57 CON: NSC	\blacktriangleright Used portable gas analyser to test metabolic variables on outdoor track \blacktriangleright In the INT group the Δ RE was moderately correlated ($r = 0.55^*$) to their Δ hopping power
Spurrs et al. (2003) [16]	INT: 8 M CON: 9 M ET	INT + CON: TV: 60-80 km·wk [range] INT: AC: 57.6 \pm 7.7 CON: 57.8 \pm 5.4	INT: 6 wks as jumps, bounds, hops in horizontal & vertical planes Progressive overload based on number of foot contacts per session Weeks 1-3: 2/day/wk Weeks 4-6: 3/day/wk CON: Continued normal ET	$\gg k_{MTU}$ determined from oscillation pattern based on equation for motion of dampened spring [87] Loads of 25, 50, 75% of the MVIC with ankle angle set at 90°	INT: \uparrow 14.9%* (left leg) \uparrow 10.9%* (right leg) at 75% MVIC load CON: NSC	\gg RE: INT: SS*** \uparrow 6.7%, 6.4%, 4.1% @ 12, 14 & 16 $km \cdot h^{-1}$ CON: NSC \gg 3km Time trial: INT: SS* 2.7% \uparrow CON: NSC	\blacktriangleright Measured both legs separately \blacktriangleright Only the INT group showed a SS** \uparrow in countermovement jump and horizontal broad jump by 13.2% and 7.8% respectively \blacktriangleright SS* large negative correlation ($r = -0.53$) between Δ broad-jump and Δ RE at 12 $km \cdot h^{-1}$

INT = intervention group, CON = control group, M = male, F = female, ET = non-specified endurance trained, REC = recreationally trained, MDT = middle distance trained (800-5000 m), LDT = long distance trained (5000 m +), TV = reported training volume at time of study, AC = aerobic capacity ($ml \cdot kg^{-1} \cdot min^{-1}$), PB = personal best race time, HRT = heavy resistance training, PLY = plyometric training, ISO = isometric training, k_{MTU} = stiffness of muscle-tendon unit, k_{AT} = Achilles tendon, k_{leg} = leg stiffness, k_{vert} = vertical stiffness, MVIC = maximal voluntary isometric contraction, DYNO = isokinetic dynamometer, 5-H = 5 consecutive maximal hops, RE = running economy, O_c = oxygen cost, E_c = total energy cost to run per unit of distance.

Data are mean \pm SD unless indicated, \uparrow - improved, N/R = study did not report
SS = statistically significant at the following level: * $P \leq 0.05$, ** $P \leq 0.01$, or*** $P \leq 0.001$.
ES = Effect size, CL = 90% confidence limits, NSC = non-significant change

3.5.2.4. *Summary of stiffness alterations*

Resistance training may enhance lower-body stiffness, and ultimately performance, via numerous mechanisms. Potential neural adaptations may be one such mechanism, which occur via enhanced motor-unit recruitment and coordination [130]. Research supports the linkage between changes in stiffness and RE, and it appears longer training periods may be required to elicit such changes at the component level [26, 129, 133, 134]. It seems research has addressed the somewhat misjudged concern of runners and coaches that strength training will lead to increased rates hypertrophy and added body mass may negatively affect performance. Storen et al. [130] along with others [14, 26], reported no significant changes in body weight after the HST, nor any harmful effects on $\dot{V}O_{2\text{peak}}$ or RE. There are inherent difficulties in drawing conclusions between results, owing to the rarity of research implementing reliable stiffness assessment techniques that are time effective, practical and valid to high performance sports settings. The optimal modality, frequency and intensity of training interventions to alter stiffness and cause positive performance changes remains elusive.

3.6. *Limitations and Future Directions*

The determinants of successful middle-distance running are multifaceted, with many mechanical variables reported to partially explain observed inter-individual differences in performance. The relationship of non-modifiable anthropometric influences on RE has been recently reviewed elsewhere [12], with several of these central in determining stiffness. One such example, resting leg length (L_0), contributes to the angular inertia and the energy cost to move the legs while running [135], and is used in the global calculation of k_{leg} [21]. Furthermore, Scholz et al. [136] proposed the amount of energy stored in the AT during running relies on the length of the moment arm (MA) of the ankle, with the amount of stored potential energy increasing with a shorter MA. This study [136] reported external 2D measures of MA explained 58% of the variance in RE (at $16 \text{ km}\cdot\text{hr}^{-1}$) across 15 highly-trained runners. Therefore, anthropometric qualities which may predispose athletes to have greater running ability, are indeed part of the scope of exploring lower-body stiffness.

When estimating energy demands of overground running, treadmill tests often set a 1% gradient to account for a lack of air resistance, as compared to overground running [137]. While making this adjustment may enhance the validity of metabolic cost, an incline may

have an effect of the calculated k_{leg} , though no research to date has investigated such influences. Furthermore, assessments of k_{leg} performed on a treadmill can be specific to the compressibility of that machine [123, 138]. Belt configuration and the rigidity of the supporting frame are considered negligible leg compression estimates, despite evidence showing surfaces with difference compliance have an influence on kinematics [139] and lower-body stiffness [138] during locomotion. Therefore, the testing surface along with its gradient should be acknowledged and more importantly, held consistent were possible.

Though longitudinal data supports RE and stiffness are likely to change together, it remains difficult to quantify how training load exposure across a runner's career (volume, frequency and intensity) or genetic predisposition (non-modifiable anthropometric qualities) affect stiffness trainability. Published differences of the physiological [140] and mechanical [141, 142] differences between African and European runners may provide some insight. These genetic and training influences are not yet fully understood and are likely to regulate the stimulus necessary to elicit stiffness adaptations and may inform an athletes ceiling for improvement. Such suggestions warrant further investigation.

Studies associating stiffness to running performance that is quantified by race [34, 126] or time-trial [16], contain uncontrolled variables that comprise a personal best time in any distance running event. For example, motivation, pacing strategies, the level of competition, and environmental conditions. Therefore, assessment of races or time-trial to compare 'ability' between runners may be restricted, despite the obviously practical advantage over metabolic testing. An additional limitation when comparing studies lies in the training status of the runners during the period the study is conducted [10] (e.g. off-season versus in season). Typically, the timing within the season is not described by authors, making comparisons of stiffness values problematic. Stiffness reported during a build-up phase (i.e. higher volume), may not be representative of what stiffness is optimal during actual completion.

Despite examining stiffness in runners, the complex relationship between global and component stiffness remains unclear. To the author's knowledge, biomechanical investigations into the adjustments of stiffness at the individual joint level are yet to be assessed following a chronic non-running training modality in highly-trained athletes. As such, it is yet to be quantified if an increase or decrease in stiffness at the hip, knee or ankle joint would be beneficial to performance [143]. Significantly, assessments of global

measures across a variety of tasks, (hopping and running) in the absence of a structural measure, make it difficult to understand how stiffness is underpinned with the body segments. Additionally, limited research exists on the acute effects of altering stiffness via external loading to the body [13]. The effect of altering a runner's load through added weights or artificially by adjusting gravity may be worthwhile avenues for future research. To further understand structures that influence running *in vivo*, agreement should be reached on a common assessment method, to understand the upper and lower limits of optimal stiffness for a given task.

3.7. Conclusions

In conclusion, literature currently supports evidence for higher stiffness associated with greater running performance across an array of assessment methods of both global and component stiffness discussed within this review. Inherent differences between methods and the links between global stiffness and MTUs are yet to be completely defined. By coupling the assessment of both metabolic and mechanical determinants of running performance, a further appreciation for the intuitive link between the two capacities can be informed. Accordingly, this will aid practitioners to understand the profile of each athlete and how their subsequent training prescription should target measurable changes to performance.

CHAPTER 4

METHODS

4.1. Research Design and outline

This study used a cross-sectional experimental design where measurements took place during the competition period of the athletics season. This approach enabled the evaluation of athletes in the same relative phase of their training (in-season). Participants were required to visit the laboratory on two occasions separated by 24 hours rest (*Figure 4-1*). Tests were performed on an indoor 60 m synthetic track and in the Strength and Conditioning and Sports Physiology laboratories at the Sports Performance Research Institute New Zealand, AUT Millennium Campus. The laboratories were well-ventilated and temperature controlled, ensuring consistent environmental conditions during testing (20.1 ± 1.8 °C and $60 \pm 7\%$ for temperature and relative humidity respectively).

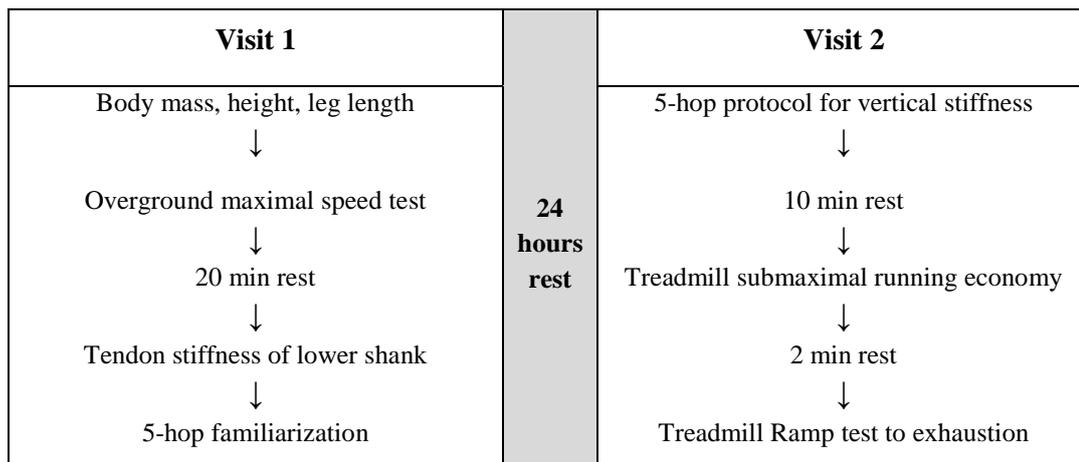


Figure 4-1. Timeline for study.

4.2. Participants

Eleven highly-trained male MD runners were recruited from local athletic clubs throughout the greater Auckland region. Participant characteristics are presented in *Table 4-1*. Requirements to participate ensured none of the runners had any musculoskeletal injuries at the time of the study nor in the six months prior to commencement. To ensure the ‘highly-trained’ classification for this sample was achieved, participants were eligible to take part if they met a minimum performance standard. This standard was defined as \geq 750 points from the most recent edition of the IAAF Scoring Tables of Athletics [144] for times in the 800 m to 5000 m events achieved in the current or previous athletic season

(see *Appendix 1* for details). These standards were set to ensure all participants were of a standard whereby they would be eligible to compete in the national track and field championships [145]. In addition, participants were required to have commenced regular training at least two years prior to participation in the study and were regularly training ≥ 5 days per week at the time of testing.

Participants were informed verbally and in written form of the risks associated with the testing and the requirements of participation. Participants were afforded the opportunity to have any questions answered and prior to commencement and provided written informed consent in accordance with the Research Ethics Committee at Auckland University of Technology, Auckland, New Zealand (Application 14/324, see *Appendix 3*).

Table 4-1. Participant characteristics (N=11).

Measure	Mean \pm SD
Age (y)	20.0 \pm 2.9
Body Mass (kg)	68.2 \pm 6.7
Height (m)	1.82 \pm 0.05
Standing Leg Length (m)	0.94 \pm 0.06
Best Race Score ¹	850 \pm 90
Best 1500 m Time (min:sec)	4:02 \pm 0:06
Weekly running volume (km \cdot wk ⁻¹)	90.8 \pm 15.6

¹ = IAAF [144] scores from the 800 m – 5000 m events

4.3 Methods

Participants were asked to arrive in a rested and hydrated state before the testing sessions, having eaten at least 2 hours prior, abstained from caffeine and avoided strenuous exercise in the preceding 24 hours. Participants were asked to bring their own racing spikes for the track sprints (visit 1), and light-weight racing shoes or flats for the treadmill and hopping assessments (visit 2).

4.3.1. Visit 1: Anthropometric measurements.

The first visit involved measurement of height, body mass (in clothes to be worn in running tests, without shoes) and resting leg length (L_0) measured as the height of the greater trochanter from the floor when standing without shoes.

4.3.2. Visit 1: Maximal velocity and leg stiffness.

The purpose of the sprint test was to determine the maximal speed each athlete could attain overground on a synthetic track surface (indoors), and functional leg spring stiffness while performing this task. Prior to testing, participants performed a standardized warm-up consisting of 10 minutes slow jogging on an outdoor track at a self-selected speed, followed by their normal warm-up stretches as if preparing for a race. This was followed by three build up runs or 'strides' over 50 m, each of increasing intensity with three to four minutes rest between strides. The experimental trials consisted of three maximal 50 m sprints. During each trial, participants were provided verbal encouragement, and were instructed to build up to their v_{\max} prior to the 40 m mark, following a similar procedure reported previously [134]. Each trial was separated by four minutes of passive rest.

Maximal sprint velocity was measured using a radar device (Stalker ATS II, Applied Concepts, Dallas, TX, USA), which was secured via a bracket adapter to a tripod positioned 2 m behind the starting line and 1 m above the ground to correspond approximately to the participant's centre of mass [146]. The radar device measured forward sprinting velocity at a frequency of 46.9 Hz and was operated remotely via laptop. The validity of this device has been demonstrated when compared to photoelectric cells in previous studies examining sprint running performance including measures of v_{\max} [37, 38].

In conjunction with the radar to measure v_{\max} , a series of ground-based photoelectric cells, (OptoJump, Microgate, Bolzano, Italy) collected step temporal characteristics between 40 - 47 m in each trial. The OptoJump system consisted of seven 1 m bars, positioned in parallel on the synthetic athletic track, allowing for normal athlete-surface interaction (see *Figure 4-2*). Each bar comprises 32 infrared light emitting diodes (LED), resulting in a system accuracy of 0.031 m at a sampling frequency of 1000 Hz. Optojump bars were connected to a laptop computer and the proprietary software (Optojump software, version 1.5.1.0) allowed for temporal measurement with a precision of 0.001 s. This system

allows simplified capture and analysis of these variables over multiple steps with excellent reliability (ICC = 0.98) [5]. In pilot testing using the radar device, MD runners commonly attained v_{\max} during this seven metre portion (40 – 47 m) of a maximal sprint.



Figure 4-2. Optojump (yellow bars) positioned on the synthetic running track to capture spatiotemporal measures during maximal overground sprinting.

4.3.3. Visit 1: Tendon mechanical properties

After the maximal sprint assessment, 20 minutes of passive rest was given before participants began the second test of visit one. Assessing maximal voluntary isometric contractions (MVICs) of the ankle plantar flexors, enabled the calculation of tendon-unit stiffness (k_T). This protocol was conducted using a Humac Norm isokinetic dynamometer (Lumex, Ronkonkoma, NY, USA), using the standardized set up in the manufacturers instructions for a plantarflexion (PF) task. Participants were positioned using a protocol described previously by Fletcher, Esau and MacIntosh [17]; where participants lay prone on the dynamometer table, with their knees fully extended (180°) and the right ankle in the anatomical neutral position (90°). The unshod right foot was fixed to the dynamometer footplate attachment using multiple Velcro straps. The ankle joint axis of rotation, defined as the medial malleolus, was aligned with the dynamometer axis of rotation. Gravity adjustments were then made by determining the shank mass supported by the ankle

attachment and the passive muscle tension using the Humac software (Lumex, Ronkonkoma, NY, USA).

Three isometric PF familiarisation trials of increasing intensity (self-perceived 50, 70 and 90% of maximum) were performed, followed by the MVIC trials. The verbal instructions to participants for the MVICs were to ramp slowly up to, and then hold their maximal effort, pushing as hard as possible against the footplate with the forefoot, while continuing to maintain heel contact with the dynamometer's footplate. The total duration of each PF trial was three seconds. During maximal trials, participants were given strong verbal encouragement to maximise consistent effort. Movement about the ankle joint was minimised by applying high tension on the Velcro strapping. Two minutes rest between each PF was observed, with five minutes between familiarization and the recorded MVICs. All data was sampled at 100 Hz using custom-built LabVIEW program (Build version: 11.0, National Instruments Corp, Austin, TX, USA) taken direction via the auxiliary outputs from the Humac Norm.

During all PF trials described above, a linear array (12L-RS, 13 MHz) ultrasound transducer (GE Healthcare, Vivid S5, USA), was used to image the right triceps surae (TS) and Achilles tendon (AT) aponeurosis, as it moved in the sagittal plane. The ultrasound probe was placed where the fascicles of the gastrocnemius (GM) meet the AT, namely the neuromuscular junction. The probe was secured to the leg using a custom-built Styrofoam cast (*Figure 4-3*) similar to previous investigations [17]. Two-dimensional (2D) video was captured at 40 Hz and later downloaded as a video file from the ultrasound unit.

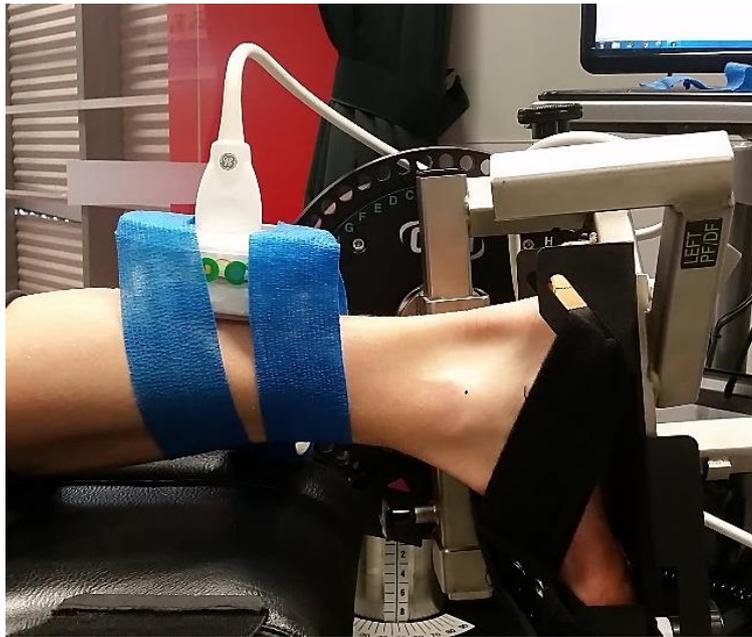


Figure 4-3. Placement of the ultrasound probe attachment with custom-built housing secured to the right shank.

The moment arm (MA) of the ankle was estimated based on a hybrid method previously described [147]. The purpose of attaining this measure was to estimate torque produced during PF at the ankle joint [147]. To gain an external measure of the ankle MA, a digitised 2D image of the medial aspect of the participant's ankle was captured against a reference tape (*Figure 4-4*), adapted from a previously reported method [136]. Using video analysis software (Kinovea 0.8.15), the distance from the medial malleolus to the most posterior part of the ankle was recorded. The measure of the MA was then corrected by subtracting the distance between the skin and the midline of the AT *in vivo* so as to represent more accurately the lever arm during ankle joint rotation [147]. This distance was estimated from a digitised ultrasound image of the AT at rest, captured with the probe placed in line with the medial malleolus while the participant lay prone (*Figure 4-5a*). Using this digitized image (*Figure 4-5b*), the measured distance from the skin to the midline of the AT (*Figure 4-5b*, white dash line), was subtracted from the original the externally visualised MA (*Figure 4-4*). This resulted in an estimate of 'true' moment arm of the ankle to be used in further analysis and calculation of stiffness (see 4.4.2).



Figure 4-4. Image of the medial aspect of the right foot placed alongside a reference tape to estimate external moment arm (indicated by the solid line).

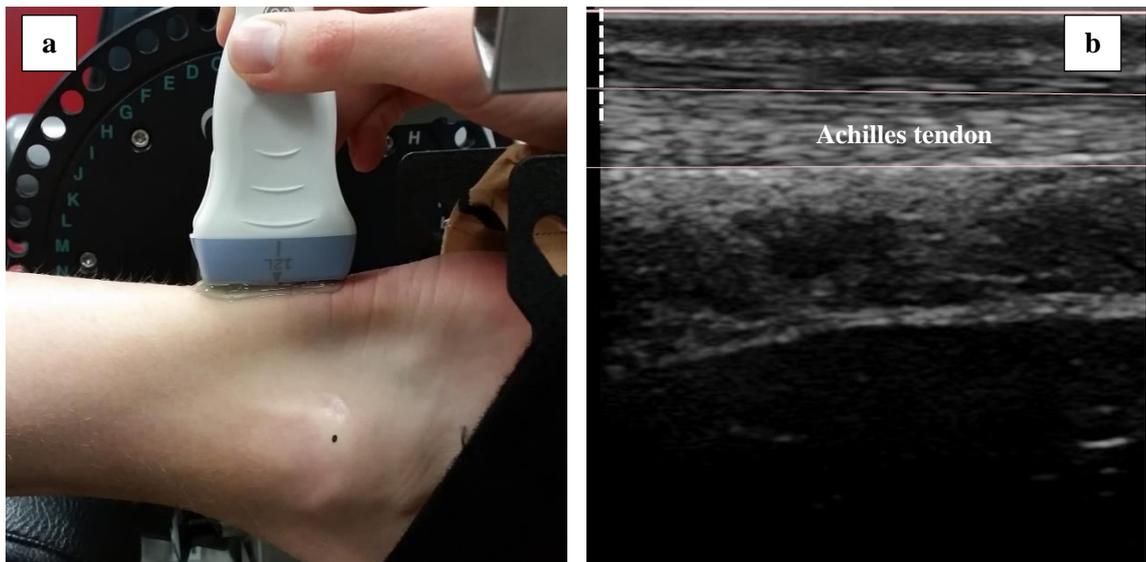


Figure 4-5: a) Position of the ultrasound probe in line with the medial malleolus;
 b) Ultrasound image captured for the measurement of the distance between the skin and the AT midline (indicated by the dashed line).

4.3.4. Visit 1: Familiarization of maximal hopping protocol

At the conclusion of visit one, participants were given instructions on the hopping task that would follow in visit 2 (see 4.3.4). Each participant performed several familiarisation trials with verbal feedback to ensure consistency in technique across participants. Familiarisation was included on visit one so that fatigue from the practice would not influence the reactive abilities tested in this protocol. This was based on work showing that familiarisation increased the reliability of hopping tasks [40].

4.3.5. Visit 2: Vertical stiffness via maximal hopping

Approximately 24 later, each participant returned to the laboratory having abstained from exercise between visits. The standardised warm-up in visit two was confined to five minutes on a stationary bike at a self-selected low resistance so as not to produce any neuromuscular fatigue in the lower limbs. A measure of maximal k_{vert} was included to determine stiffness via a simple field-based assessment, requiring only minimal equipment. It was previously proposed that the ability to produce a stiff rebound during sprinting can be investigated via measures of stiffness in a rebounding hopping task [37]. The assessments were performed as a series of five consecutive maximal hops (5-H) from a plyometric box onto a contact mat (SWIFT SpeedMat, Swift, Wacoi, QLD, Australia). The 5-H protocol was adapted based on a similar test used previously in distance runners to determine stiffness [16], but modified to include a drop-jump start. Specifically, the current protocol involved vertical straight-leg hops (uni- and bilateral), initiated from a box (0.32 m) as a drop jump. This drop onto the contact mat, was followed immediately by five continuous maximal hops. The drop jump start was to reduce the effect of the initial concentric energy requirement to begin the bouncing cycle from a static start. Participants were instructed to aim for maximal height while keeping contact times as short as possible and maintaining their hands on their waist. Two trials for both uni- and bilateral protocols were recorded, with the bilateral task first, and unilateral left and right following. Three minutes rest was given between each trial.

4.3.6. Visit 2: Aerobic measures of performance

Following a 10 minute, passive rest from the 5-H tasks, a treadmill test for RE and peak oxygen consumption ($\dot{V}O_{2\text{peak}}$) was performed via a continuous step protocol, beginning at 12 km·h⁻¹ at a 1.0% gradient. Each of the submaximal stages for the determination of RE were four minutes at 12, 14, 16 and 18 km·h⁻¹. The purpose of the aerobic testing protocol was to determine each participant's maximal oxygen uptake following assessment of their oxygen combustion at submaximal running velocities. Prior to testing, a metabolic gas-analyser (ParvoMedics TrueOne 2400, Salt Lake City, USA) was calibrated in accordance with the manufacturer's instructions and deemed acceptable when variations of < 1% were achieved. In addition, a heart rate monitor (Polar Electro Oy, Kempele, Finland), headpiece and pneumo mouth piece were fitted securely to the participant. Expired gases were measured every 15 seconds using a metabolic cart for determination of: oxygen consumption ($\dot{V}O_2$), carbon dioxide consumption ($\dot{V}O_2$),

minute ventilation ($\dot{V}E$), ventilatory equivalents for oxygen ($\dot{V}E/\dot{V}O_2$) and carbon dioxide ($\dot{V}E/\dot{V}CO_2$), respiratory exchange ratio (RER).

At the conclusion of the four submaximal stages, a two-minute rest was observed while the participant stood stationary on the treadmill. Following the assurance the participant was ready to continue, an incremental test was performed to volitional exhaustion to determine maximal oxygen uptake ($\dot{V}O_{2peak}$). The test began at a constant velocity of 16 $\text{km}\cdot\text{hr}^{-1}$ and a 1% gradient. The treadmill grade was then increased by $1\%\cdot\text{min}^{-1}$ until terminated voluntarily by the subject. The highest $\dot{V}O_2$ over a 15-second period during the test was considered the athlete's $\dot{V}O_{2peak}$.

4.3.7. Visit 2: Submaximal running leg stiffness

Within the submaximal aerobic running stages described above, foot strike of both the left and right foot were captured via 2D video at 210 Hz (Casio EX-FH20, Casio Computer Co., Tokyo, Japan), between the third and fourth minute of each stage. It was previously reported that after 2 - 3 minutes running on a treadmill, running characteristics are highly reproducible [148]. Using video analysis software (Kinovea 0.8.15) frame-by-frame counts were used to measure contact time (t_c), and flight time (t_f), along with the calculation of step frequency (S_f) and step length (S_L). The treadmill on which this was assessed (Saturn, h/p Cosmos, Japan), was supported by a very stiff metal frame, and it was assumed calculation of k_{leg} was not influenced by the treadmill stiffness, as no spring effect of the running surface was observed in the high-speed 2D footage.

4.4. Data Processing

4.4.1. Physiological variables

The oxygen and energy cost of running (E_{run}) was determined using data averaged over the last minute of each submaximal stage of the treadmill test, for assurance of physiological steady-state. RE was expressed as the relative $\dot{V}O_2$ required to run at 14 $\text{km}\cdot\text{h}^{-1}$, as well as the percent $\dot{V}O_{2peak}$ utilised to run at this speed. In addition, RE was also expressed as E_{run} as this expression has been reported to be less influenced by speed than the oxygen cost per min alone [112]. Shaw et al. [112] recently proposed this was due to inability of oxygen cost as a measure of RE to account for deviations in substrate metabolism. In contrast, the respiratory exchange ratio (RER) between $\dot{V}O_2$ and $\dot{V}CO_2$,

is inherent within the calculation of E_{run} . Therefore the calculation of E_{run} incorporated energy expenditure (EE), estimated using the modified Weir equation [149] (*Table 4-2* Eq. 1), and subsequently converted into E_{run} relative to both body mass and distance covered ($\text{kcal}\cdot\text{kg}\cdot\text{km}^{-1}$).

A further measure of running performance is the velocity at which $\dot{V}O_{2peak}$ occurs ($v\text{-}\dot{V}O_{2peak}$). As previously applied in the literature [113], a conversion for equivalent running velocity (v) on the flat was calculated from the treadmill incline when $\dot{V}O_{2peak}$ was reached based on the publication by Brooks Fahey and White [150] (*Table 4-2* Eq. 3).

Table 4-2. Summary of calculations used from treadmill assessment

Eq.	Variable	Calculation
(1)	Energy Expenditure (EE) ($\text{kcal}\cdot\text{min}^{-1}$) [149]	$EE = 3.941 \times \dot{V}O_2(l \cdot \text{min}^{-1}) + 1.106 \times \dot{V}CO_2(l \cdot \text{min}^{-1})$
(2)	Energy cost of running (E_{run}) ($\text{kcal}\cdot\text{kg}\cdot\text{km}^{-1}$)	$E_{run} \left(\frac{EE (\text{kcal} \cdot \text{min}^{-1})}{\text{Body Mass (kg)}} \right) \times \text{running velocity (min} \cdot \text{km}^{-1})$
(3)	Velocity (v) from treadmill incline (i)	$v = v_{\text{treadmill}} + (v_{\text{treadmill}} \times 0.045) \times i \%$

4.4.2. Biomechanical calculations

During the assessment of overground sprinting, the trial eliciting the highest v_{max} was used for further comparative analysis and leg stiffness calculation (*Table 4-3*, Eq. 5). The data captured from the radar was imported into the manufacturer's software (Stalker ATS II, Applied Concepts, Dallas, TX, USA) and cropped of any data points where the participants were stationary. Outlier points showing zero velocity were then deleted and exported to Excel for subsequent analysis. Additionally, using the Optojump software (Optojump software, version 1.5.1.0) the mean temporal measurements were calculated from all steps (mean of left and right limbs) that occurred within the 7 m capture zone. The remaining data was used in the calculation of k_{leg} using a previously established reliable method [21] (*Table 4-3*, Eq. 5).

For the calculation of k_T , ultrasound video files were imported into video analysis software (Kinovea version 0.8.15) and calibrated using the ultrasound depth scale embedded in the video file. A point on the video image where a clearly visible GM fascicle inserted into the deep aponeurosis of the AT was identified throughout each contraction (*Figure 4-6*). The maximal displacement of this fascicle–aponeurosis junction in each PF trial was interpreted as tendon elongation. The force produced by the entire muscle-tendon unit (MTU) was calculated by dividing the ankle joint MA by the MTU moment from the dynamometer output. The raw data from the Humac was exported into a spreadsheet with peak force within each trial used for the MTU moment calculation. With these two variables (elongation and force), stiffness for each MVIC trial was calculated as force divided by length change (*Table 4-3*, Eq. 6). The k_T for each participant was then represented as the mean of the three recorded MVIC trials. This representation of stiffness was used in subsequent analysis, as we found it showed the best overall reliability from testing in our laboratory (ICC = 0.87) when compared to other expressions such as the slope of the force-elongation curve. Further details on how this measure's reliability was determined and compared to an alternate stiffness expression are presented in *Appendix 2*. Based on a lack clear ultrasound images attained for one participant, he was removed from the further k_T analysis.

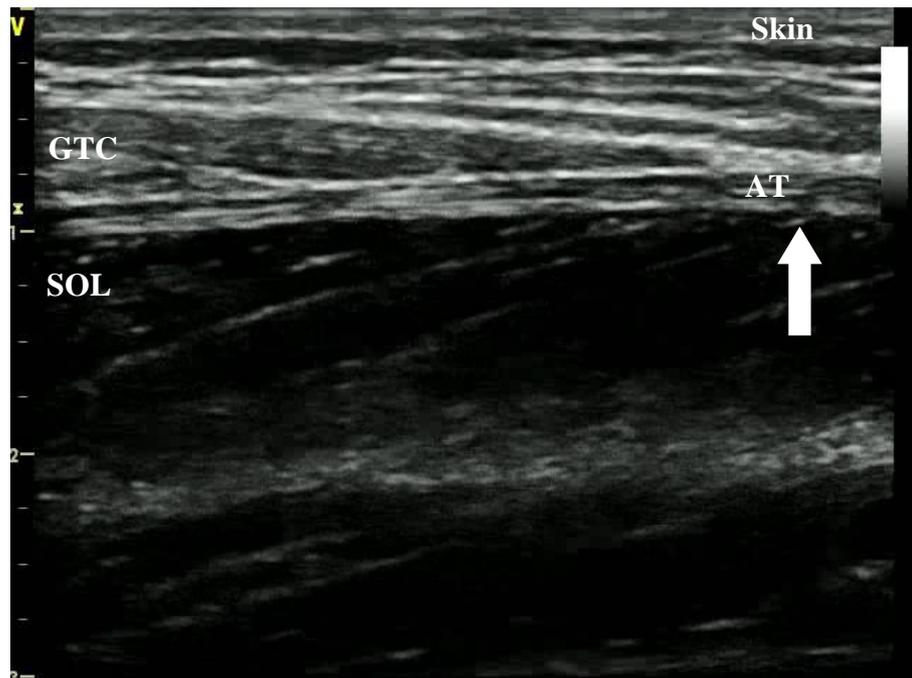


Figure 4-6. Ultrasound image of the right medial gastrocnemius (GTC) and soleus (SOL) as they converge to form the Achilles tendon (AT). White stripes represent echoes generated by collagen-rich tissue surrounding the echo-absorptive fascicles. The white arrow indicates an identifiable landmark on the AT aponeurosis that was seen in multiple images as it moves more proximally (to the left in this figure) and was used to calculate tendon elongation during each trial.

From the 5-H test, t_c and t_f of the middle three hops were averaged to calculate k_{vert} (Table 4-3, Eq. 4), with the trial eliciting the greatest k_{vert} across each task (bilateral, left and right 5-H) used in further analysis. By excluding the first contact and flight phase, the influence of the variability of initial drop landings was minimised.

Finally, the calculation of k_{leg} from the submaximal stages of the treadmill assessment used the average t_c and t_f of 20 consecutive steps captured in the third minute of each stage. This method was previously reported to produce reliable estimates of t_c and t_f for the calculation of k_{leg} [21, 151] (Table 4-3, Eq. 5).

Table 4-3. Summary of biomechanical stiffness models used and their method of data collection in this study

Eq.	Model [reference]	Task (Equipment)	Stiffness calculation
(4)	Vertical stiffness [21]	5 continuous unilateral & bilateral hops from a drop-jump start [32 cm] (box, contact mat)	$k_{\text{vert/time}} = \frac{F_{\text{max}}}{\Delta y}$ $F_{\text{max}} = mg \frac{\pi}{2} \left(\frac{t_f}{t_c} + 1 \right)$ $\Delta y = -\frac{F_{\text{max}}}{m} \frac{t_c^2}{\pi^2} + g \frac{t_c^2}{8}$
(5)	Leg stiffness [21]	Maximal 50 m sprint (Optotrack, radar gun) Submaximal Treadmill running (high-speed camera)	$k_{\text{leg/time}} = \frac{F_{\text{max}}}{\Delta L}$ $F_{\text{max}} = mg \frac{\pi}{2} \left(\frac{t_f}{t_c} + 1 \right)$ $\Delta L = L_0 - \sqrt{L_0^2 - \left(\frac{vt_c}{2} \right)^2} + \Delta y$ $\Delta y = -\frac{F_{\text{max}}}{m} \frac{t_c^2}{\pi^2} + g \frac{t_c^2}{8}$
(6)	Muscle-Tendon Unit Stiffness	Ankle plantarflexion (Humac Norm, ultrasound)	$k_T = \frac{F_{\text{MTU}}}{\text{AT } \Delta L}$ $F_{\text{MTU}} = \frac{\text{Force}}{MA}$

k_{vert} = vertical stiffness, k_{leg} = leg stiffness, k_T = Achilles tendon-unit stiffness F_{max} = peak vertical force, Δy = centre of mass displacement, m = subject mass, t_c = contact time, t_f = flight time, g = acceleration due to gravity, ΔL = change in leg length, L_0 = trochanterian height, v = horizontal velocity, F_{MTU} = muscle-tendon unit force, $\text{AT } \Delta L$ = maximal Achilles tendon-unit elongating.

4.4.3. Statistical analysis

Means and standard deviations were calculated for all variables, along with the between-participant coefficients of variation (CV%) for RE and $\dot{V}O_{2\text{peak}}$ to determine the homogenous nature of this group of participants. Associations between each measure of stiffness and between stiffness measures and running performance indicators were investigated by calculating Pearson's product moment correlation coefficients in SPSS (IBM SPSS version 20.0, Chicago, IL, USA). The magnitude of correlations were interpreted based on the following scale: *trivial* (0.0-0.1), *small* (0.1-0.3), *moderate* (0.3-0.5), *large* (0.5-0.7), *very large* (0.7-0.9), or *extremely large* (0.9-1.0) [105]. In addition, ninety percent confidence limits for all correlations and magnitude based inferences were

then derived using an Excel spreadsheet [152]. Based on a 2 SD change in stiffness (moving from typically low to typically high stiffness), a correlation of 0.1 was used to give the smallest worthwhile change in RE of 0.6% [102]. Where the correlation had a >5% probability of being substantially positive and a >5% probability of being substantially negative the inference was stated as '*unclear*' [153, 154]. Otherwise the outcome was clear and the inference was based on the likelihood the true value of the correlation was greater than 0.1 using the following scale: 25-75%, possibly; >75%, likely; >95%, very likely; >99.5%, most likely [155]. Similarly to a previous study [5], the Pearson's correlation coefficients (r) were squared resulting in a percentage of variance explained (%) between the performance measures and stiffness characteristics.

CHAPTER 5

RESULTS

5.1. *Physiological and biomechanical characteristics*

The characteristics determined from the various physiological and biomechanical tests are presented in *Table 5-1*. The high training level of the participants was recognised by their range of $\dot{V}O_{2\text{peak}}$ values (62.6 - 73.2 ml·kg·min⁻¹). Additionally, sample homogeneity was evident by the low between athlete coefficient of variation for $\dot{V}O_{2\text{peak}}$ (5.6%), percent of $\dot{V}O_{2\text{peak}}$ required to run at 14 km·hr⁻¹ (4.6%) and RE when expressed as either energy cost to run per kilometre (2.7%) or relative oxygen cost per min (3.2%).

Examining the relationships between stiffness and RE (expressed as E_{run}), the largest observed association was a *very large* negative correlation between RE and stiffness in the Achilles tendon unit ($r = -0.703$), as shown in *Figure 5-1*. A full summary of the relationships between various lower-body stiffness and running performance measures (RE and v_{max}) are presented in *Table 5-2*. Furthermore, the association between the measured ankle moment arm on the right shank with RE and v_{max} was *unclear*.

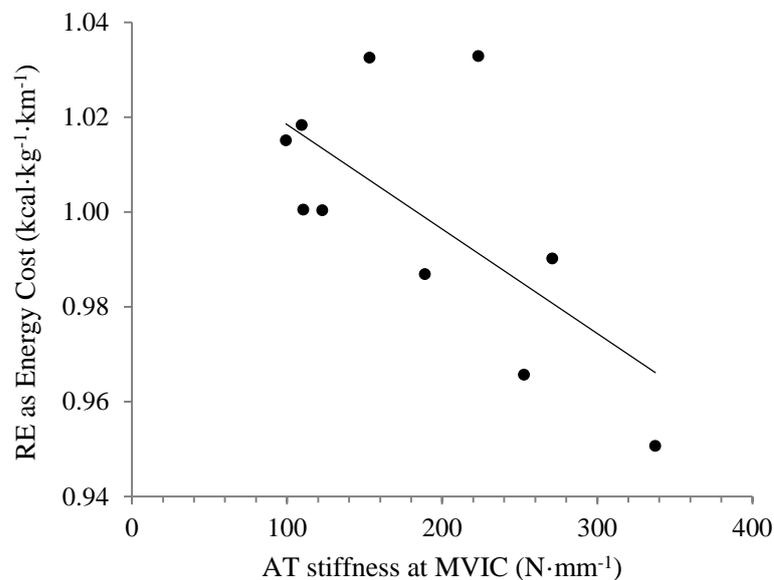


Figure 5-1. The relationship between the Achilles-tendon (AT) stiffness and running economy (RE).

Table 5-1. Participant characteristics for key physiological and biomechanical variables (N=11[♦]).

Measure	Mean ± SD
<i>Metabolic</i>	
$\dot{V}O_{2peak}$ (ml·kg·min ⁻¹)	67.6 ± 3.8
$v \cdot \dot{V}O_{2peak}$ (km·hr ⁻¹)	20.8 ± 1.0
RE as O ₂ cost at 14 km·hr ⁻¹ (ml·kg·min ⁻¹)	46.8 ± 1.5
RE at 14 km·hr ⁻¹ (% $\dot{V}O_{2peak}$)	68.9 ± 3.1
RE as E _{run} (kcal·kg ⁻¹ ·km ⁻¹)	1.00 ± 0.03
<i>Biomechanical</i> ^{S-S}	
step frequency (steps·min ⁻¹)	166.6 ± 4.6
step length (m)	1.40 ± 0.04
contact time (s)	0.239 ± 0.011
flight time (s)	0.121 ± 0.015
<i>Biomechanical</i> ^{M-R}	
v_{max} (m·s ⁻¹)	8.66 ± 0.46
step frequency (steps·min ⁻¹)	246.7 ± 12.2
step length (m)	2.10 ± 0.11
contact time (s)	0.127 ± 0.011
flight time (s)	0.114 ± 0.007
<i>Vertical Stiffness</i>	
k_{vert} - bilateral 5-H (kN·m ⁻¹)	20.63 ± 5.25
k_{vert} - right unilateral 5-H (kN·m ⁻¹)	9.76 ± 6.79
k_{vert} - left unilateral 5-H (kN·m ⁻¹)	7.92 ± 1.99
<i>Leg Stiffness</i>	
^{S-S} k_{leg} (kN·m ⁻¹)	4.90 ± 1.03
^{M-R} k_{leg} (kN·m ⁻¹)	8.40 ± 2.26
<i>Muscle-Tendon Properties</i> [♦]	
MVIC peak moment (N·m ⁻¹)	107.5 ± 43.1
k_T at MVIC (N·mm ⁻¹)	186.9 ± 81.6

[♦]N=10 for muscle-tendon analysis, RE = running economy, E_{run} = energy cost to run, S-S = submaximal steady-state treadmill running at 14 km·hr⁻¹, M-R = maximal overground running, v_{max} = athletes maximal velocity, k_{vert} = vertical stiffness, k_{leg} = leg stiffness, k_T = Achilles tendon-unit stiffness, MVIC = maximal voluntary isometric contraction, 5-H = continuous 5-hop plyometric assessment

Table 5-2. Correlations between lower-body stiffness, running economy and sprint performance characteristics in male MD runners (N=11[♦]).

	Running Economy (kcal·kg ⁻¹ ·km ⁻¹)		Maximal sprint velocity (m·s ⁻¹)	
	Correlation ^a ± 90% CL	Magnitude based inference	Correlation ^a ± 90% CL	Magnitude based inference
<i>Vertical Stiffness</i>				
Bilateral 5-H k_{vert} (kN·m ⁻¹)	-0.42 ±0.45	<i>Unclear</i>	0.27 ±0.50	<i>Unclear</i>
Right Leg 5-H k_{vert} (kN·m ⁻¹)	-0.60 ±0.37	<i>Large</i> ***	0.44 ±0.45	<i>Unclear</i>
Left Leg 5-H k_{vert} (kN·m ⁻¹)	-0.46 ±0.44	<i>Moderate</i> **	0.49 ±0.43	<i>Moderate</i> **
<i>Leg Stiffness</i>				
<i>S-S</i> k_{leg} (kN·m ⁻¹)	-0.25 ±0.50	<i>Unclear</i>	0.18 ±0.51	<i>Unclear</i>
<i>M-R</i> k_{leg} (kN·m ⁻¹)	-0.63 ±0.35	<i>Large</i> ***	0.59 ±0.38	<i>Large</i> **
<i>Muscle-Tendon Properties</i> [♦]				
k_{T} at MVIC (N·mm ⁻¹)	-0.70 ±0.33	<i>Very Large</i> ***	0.52 ±0.42	<i>Large</i> **
MVIC peak moment (N·m ⁻¹)	-0.50 ±0.42	<i>Large</i> **	0.40 ±0.46	<i>Unclear</i>

[♦] N = 10 for Muscle-tendon assessment, *S-S* = submaximal steady-state treadmill running at 14 km·hr⁻¹, *M-R* = maximal overground running, k_{vert} = vertical stiffness, k_{leg} = leg stiffness, 5-H = 5-hop plyometric assessment MVIC = maximal voluntary isometric contraction, k_{T} = Achilles tendon-unit stiffness

^a = Correlation magnitude thresholds: 0.1 = *small*; 0.3 = *moderate*; 0.5 = *large*; 0.7 = *very large*; 0.9 = *extremely large*, and their inverse. Asterisks indicate effects clear at the 90% level and likelihood that the relationship is substantial, as follows: ****most likely, ***very likely, **likely, *possible.

In the present cohort, there was a *large* positive correlation with RE and Sf , along with the reciprocal *large* negative relationship between RE and SL (Table 5-3). The temporal measures of t_f and t_c had an *unclear* relationship with RE while mean t_c during sprinting had a *very large* correlation to overground v_{max} .

Table 5-3. Correlations between spatiotemporal measures and key running performance characteristics in male distance runners (N=11).

Measures [#]	Running Economy (RE) (kcal·kg ⁻¹ ·km ⁻¹)		Maximal sprint velocity (v_{max}) (m·s ⁻¹)	
	Correlation ^a ± 90% CL	Magnitude based inference	Correlation ^a ± 90% CL	Magnitude based inference
Sf (steps·min ⁻¹)	0.63 ±0.35	<i>Large</i> ***	0.40 ±0.46	<i>Unclear</i>
SL (m)	-0.64 ±0.35	<i>Large</i> ***	0.59 ±0.38	<i>Large</i> **
t_c (s)	-0.21 ±0.51	<i>Unclear</i>	-0.75 ±0.27	<i>Very large</i> ***
t_f (s)	-0.26 ±0.50	<i>Unclear</i>	0.15 ±0.52	<i>Unclear</i>

[#] = RE correlations from submaximal steady-state treadmill running at 14 km·hr⁻¹

Sf = step frequency, SL = step length, t_c = contact time, t_f = flight time, v_{max} measures from maximal overground running.

^a = Correlation magnitude thresholds: 0.1 = *small*; 0.3 = *moderate*; 0.5 = *large*; 0.7 = *very large*; 0.9 = *extremely large*, and their inverse. Asterisks indicate effects clear at the 90% level and likelihood that the relationship is substantial, as follows: ****most likely, ***very likely, **likely, *possible.

5.2. Comparisons of stiffness measurements

When comparing measurements of stiffness with each other, the most significant result was the observed correlation between the k_{vert} during a unilateral 5-H test on the right leg, and k_{leg} during overground maximal sprinting was *extremely large* ($r = 0.92$). Table 5-4 summarises additional correlations between the methods of representing stiffness.

Table 5-4. Correlation matrix of relationships between methods of assessing lower body stiffness in male distance runners (N=11[♦])

Method	Task ^a	1. Run ^{M-R}	2. Run ^{S-S}	3. 5-H BL	4. 5-H UL	5. PF ^{T-S}
Run (<i>k</i> _{leg})	1. Run ^{M-R}		<i>V. large</i> **** 0.77 ±0.25	<i>V. large</i> *** 0.75 ±0.27	<i>Ex. Large</i> **** 0.92 ±0.1	<i>Very Large</i> *** 0.73 ±0.31
	2. Run ^{S-S}	<i>V. large</i> **** 0.77 ±0.25		<i>Large</i> *** 0.60 ±0.37	<i>Large</i> *** 0.67 ±0.33	<i>Large</i> *** 0.69 ±0.34
Hopping (<i>k</i> _{vert})	3. 5-H BL	<i>V. large</i> *** 0.73 ±0.27	<i>Large</i> *** 0.60 ±0.37		<i>V. large</i> *** 0.74 ± 0.28	<i>Unclear</i> 0.27 ±0.52
	4. 5-H UL	<i>Ex. Large</i> **** 0.92 ±0.1	<i>Large</i> *** 0.67 ±0.33	<i>V. large</i> *** 0.74 ± 0.28		<i>Large</i> *** 0.63 ±0.38
Isometric [♦] (<i>k</i> _T)	5. PF ^{T-S}	<i>Very Large</i> *** 0.73 ±0.31	<i>Large</i> *** 0.69 ±0.34	<i>Unclear</i> 0.27 ±0.52	<i>Large</i> *** 0.63 ±0.38	

[♦]N = 10 for Muscle-tendon assessment, *S-S* = submaximal steady-state treadmill running at 14 km·hr⁻¹, *M-R* = maximal overground running, 5-H BL = bilateral 5 maximal hopping task, 5-H UL = unilateral right leg 5 maximal hopping task, PF^{T-S} = isometric plantarflexion of the triceps surae and Achilles tendon MTU, *k*_{leg} = leg stiffness, *k*_{vert} = vertical stiffness, *k*_T = Achilles tendon-unit stiffness

^a = Correlation ± 90% CL with magnitude thresholds of: 0.1 = small; 0.3 = moderate; 0.5 = large; 0.7 = very large; 0.9 = extremely large, and their inverse. Asterisks indicate effects clear at the 90% level and likelihood that the relationship is substantial, as follows: ****most likely, ***very likely, **likely, *possible.

CHAPTER 6

DISCUSSION

The primary aim of this study was to profile lower-body stiffness, running economy and maximal velocity in highly-trained male middle-distance runners and investigate associations between these qualities. The findings suggest that MD runners with greater component stiffness in the Achilles tendon, together with greater vertical stiffness required a lower energy cost to run at a submaximal pace. Addressing the secondary aim, examining relationships between multiple stiffness measures from field and laboratory settings, revealed several significant associations. Previous studies have not examined multiple global stiffness assessments alongside *in-vivo* tendon measures within a population of runners of any kind. The MTU of the lower-limb, which influences SSC movements at the ankle joint, influences global models of lower-body stiffness in hopping and running. Both global and component stiffness have previously been linked to running performance, especially RE, across athletes with similar high-level aerobic capacity [16, 31, 156] and by comparing stiffness to race or time-trial performance [16, 34, 126]. Results from this thesis lend support to earlier findings. A paucity of research exists that explores relationships between stiffness and v_{\max} in MD runners in overground sprinting, a critical determinant of race outcomes of MD events [8]. Thus, the finding that v_{\max} showed a clear relationship with Achilles tendon stiffness in highly-trained MD runners is novel. It is not possible to directly compare all findings of the current study to that of previous research regarding associations between measures, given this is the first study to utilize a specific combination of methodologies for exploring stiffness. Regardless, where appropriate, results are evaluated alongside other studies that utilized some of these methods to examine a similar cohort of athletes.

6.1. Stiffness and performance in male middle-distance runners

The validity of the comparisons between stiffness and performance was enhanced by the homogeneity of the athletes, for example, v_{\max} ($8.66 \pm 0.46 \text{ m}\cdot\text{s}^{-1}$; CV = 5.3%) and RE as energy cost ($1.00 \pm 0.03 \text{ kcal}\cdot\text{kg}^{-1}\cdot\text{km}^{-1}$; CV = 2.7%) had a small coefficient of variation (CV). When expressing RE as the relative oxygen cost to run at $14 \text{ km}\cdot\text{h}^{-1}$ ($46.8 \pm 1.5 \text{ ml}\cdot\text{kg}\cdot\text{min}^{-1}$), the results in the current study fall into the classification of ‘highly-trained’

males from the recent review by Barnes and Kilding [12] to state normative RE values. In the current study, participants utilized $69.4 \pm 3.5\%$ of their maximal oxygen uptake during the final minute of the $14 \text{ km}\cdot\text{h}^{-1}$ stage. This speed represented an intensity at which runners of this calibre reported that a large proportion of their training was undertaken [157].

Comparing the mechanical parameters assessed in this study, the k_{AT} values of the 11 runners ($186.9 \pm 81.6 \text{ N}\cdot\text{mm}^{-1}$) were similar to those reported by Fletcher, Pfister and MacIntosh [33] in a similar recreationally-trained cohort ($191 \pm 16.6 \text{ N}\cdot\text{mm}^{-1}$). The present k_{leg} values during sprinting ($8.40 \pm 2.26 \text{ kN}\cdot\text{m}^{-1}$, $\text{CV} = 21\%$) were lower than those reported by Joseph et al. [45] ($13.18 \pm 2.96 \text{ kN}\cdot\text{m}^{-1}$, $\text{CV} = 23\%$) from a comparable cohort and the same calculation from a validated method by Morin et al. [21]. It is plausible that Joseph and colleagues' methodological differences of stiffness, expressed across a range of running velocities overground ($2.5\text{--}6.5 \text{ m}\cdot\text{s}^{-1}$), would be contributing factors for these discrepant values, as the effect of running surface and velocity on k_{leg} has been reported elsewhere [123, 138]. The k_{vert} values from the bilateral hopping task ($20.63 \pm 5.25 \text{ kN}\cdot\text{m}^{-1}$), were higher than reported in a previous study ($9.4 \pm 2.2 \text{ kN}\cdot\text{m}^{-1}$) by Barnes et al. [31], yet the present values partially overlapped k_{vert} in non-runners ($17.6\text{--}57.6 \text{ kN}\cdot\text{m}^{-1}$) presented by Dalleau et al. [51]. While Barnes et al. [31] performed a similar task to this study of five consecutive maximal hops, their assessment was initiated from standing rather than as a drop jump. Performing a drop jump start reduces effort placed on the initial concentric phase needed to begin repeated maximal hopping. However only the middle three hops were included in the data analysis, consistent with the approach of Barnes et al. [31], which would have removed effects from the drop jump start on stiffness models. A difference in the equipment used to measure flight and contact times, (force plate versus contact mat) may also explain some of the differences in values and potentially in the observed link of k_{vert} with RE. Nevertheless, Dalleau et al. [51] have reported an excellent relationship ($r = 0.98$) between force plates and contact mats during maximal repeated hopping.

6.2. Component stiffness and running performance

The existence of spring-like mechanics, originally inferred only at the level of the whole body or leg, have recently been measured in selected muscles, tendons, and ligaments that contribute to the limb's overall behaviour during movement [39]. During the last two

decades, technical progress in ultrasonography techniques have enabled a greater understanding of the structural and functional variables of the muscle–tendon unit [77]. While these methods help to establish the contribution of tendon elasticity while running [158], the muscle–tendon behaviour affording elite performance, remains undefined. The current finding that a *very large* negative correlation ($r = -0.70$) existed between component stiffness of the Achilles tendon (k_{AT}) and RE, suggested that participants with a greater stiffness required a lower energy cost (expressed as $\text{kcal}\cdot\text{kg}\cdot\text{min}^{-1}$) to run at an absolute submaximal pace. The relationship between metabolic and mechanical power in running has been presented previously [32], indicating that variables relating to muscle power (i.e. force–length–velocity relationships) partially explain differences in RE within a similar group of athletes [17, 52, 159]. The energy required to perform SSC actions in the active components of the lower body is a metabolically costly process, and thus energy stored in the passive tendon structures has potential to afford a lower aerobic energy requirement [31]. The current results support previous assertions [16, 29, 32, 33, 79] that reported greater levels of stiffness in the Achilles tendon were associated with more efficient energy transfer at submaximal running speeds.

Fletcher et al. [33] reported that female runners ($N=18$) showed a significant ($P < 0.01$) relationship between k_{AT} and RE at a range of submaximal intensities, while males ($N=11$) did not. Thus, the current results suggest that males may indeed show a negative relationship between k_{AT} and RE. The current findings however, contrast with those of Kubo et al. [126], who reported that in trained runners competing in the 5000 m event, k_{AT} was positively correlated to race time (i.e. more compliant runners had a faster race time). While their definition of performance was task specific, Kubo et al. [126] did not measure RE to make a metabolic comparison between runners. Moreover, Kubo et al. reported stiffness as the slope of the force–elongation curve and performed additional measurements to incorporate tendon thickness into their calculations. The methodological differences between the current study and that of Kubo et al. may explain the conflicting direction of this relationship between k_{AT} and performance.

While the present study used a similar ultrasound technique to that of Fletcher et al. [33], the additional corrections to account for ankle joint rotation were not implemented in final data analysis of this study. During pilot testing, we found including this added 2D correction method [33] led to a large amount of measurement noise when tracking the markings on the ankle via 2D video. This extra source of error was subsequently removed

by assuming the small amount of movement (i.e. deviation from ‘true isometric’ contraction) was insignificant to determining differences between the stiffer and compliant AT’s in runners. The MVICs in the present study were performed up to the point that participants could maintain their calcaneus in the dynamometer footplate, with the assistance of strong Velcro fixations. During analysis, we attempted to calculate stiffness values across the time-synched ultrasound and dynamometry data. However, owing to the image resolution and frame rate (40 Hz) of the ultrasound video files, accurate tracking of a fascicle-aponeurosis across the full force spectrum, could not be obtained from a single trial.

6.3. Global stiffness and running performance

6.3.1 Leg stiffness

The measurement of k_{leg} was performed under two separate conditions to further understand how this variable is represented across different velocities. There was an *unclear* relationship between k_{leg} and RE when calculated at the submaximal velocity of 14 km·hr⁻¹. Conversely, the calculation of k_{leg} when measured while sprinting revealed *large* relationships with both v_{max} ($r = 0.59$) and submaximal RE ($r = -0.63$). Higher leg-spring stiffness, as assessed when operating at maximal functional capability, is a beneficial attribute for MD runners wanting to achieve higher maximal velocities and reduce energy cost at submaximal speeds [13, 29, 31]. The conflicting results in the presented correlations between k_{leg} and performance between speeds are believed to relate to ground contact modifications between these two running speeds. The average ground contact times by runners in the current study were reduced by 47% (0.239 s to 0.127 s) when the 14 km·hr⁻¹ and maximal sprinting conditions were compared. A longer ground contact allows the leg time to generate appropriate spring-stiffness through contractile work within the MTU. However, to generate greater stiffness during maximal sprinting, a significant elastic component contribution is required due to the faster rate of force development observed within stiffer elastic components [160]. The observation of Bojsen-Moller et al. [160] is also consistent with findings of Kubo et al. [161], who found greater k_{AT} related to drop jump height with short contacts, when compared to height achieved in a countermovement jump. Together, with evidence from Arampatzis et al. [36] who stated that sprinters possessed higher k_{AT} than endurance runners, it would appear that assessments of k_{leg} under maximum efforts are likely to provide a better

indication of the elastic rather than contractile capability of a runner. Consequently, this assertion can explain why k_{leg} at slower running speeds did not relate well to submaximal RE, as longer contact times do not assess a large involvement of the elastic (tendon) energy contribution.

In the steady-state running protocol used in the current study, four submaximal speeds (from 12 to 18 km·hr⁻¹) were employed in sequence for physiological and biomechanical investigation. The relationship between k_{leg} and RE were taken at the second (14 km·hr⁻¹) stage, having allowed the participants time to familiarize themselves with the treadmill before k_{leg} was measured. Due to the time constraints of this study, runners were not requested to attend multiple laboratory visits for the prior calculation of relative aerobic speeds. By selecting 14 km·h⁻¹ for calculation of energy cost (E_C), measurements were assumed to be below the lactate threshold in athletes of this calibre [162]. Finally, as discussed in *Chapter 2*, differences between $k_{leg/dynamic}$ and $k_{leg/time}$ as used in this study, were highlighted by Arampatzis et al. [50], who compared these two expressions over increasing velocities. When applying the original model described by McMahon and Cheng [1], ΔL was significantly overestimated. Therefore, the measures of $k_{leg/time}$ in the current study were based on the model proposed by Morin et al. [21], which had greater convergent validity to the measure of $k_{leg/dynamic}$.

6.3.2. Vertical stiffness

The second assessment of the spring-mass model was the measurement of k_{vert} from a series of five maximal hops (5-H). In the present study, a *large* negative relationship ($r = -0.60$) between k_{vert} and RE was evident when the 5-H was performed unilaterally on the right leg while the left and bilateral 5-H indicated a *moderate* and an *unclear* relationship respectively. Conversely, the left limb 5-H showed a *moderate* positive ($r = 0.49$) correlation between k_{vert} and v_{max} , while the right and bilateral associations with sprint velocities were *unclear*.

Vertical hopping methods have frequently been employed as a substitute for running measurements, as they allow a time-efficient assessment of stiffness when space and/or resources are limited [4]. Accordingly, measurement of the variables to calculate k_{vert} were made via a contact mat, thus exploring a method that would appeal to practitioners who do not always have access to more expensive force plate technology. Further, an assessment of both limbs unilaterally was also undertaken, as running is a unilateral task

performed as a series of propulsive bounds. Again, due to time constraints, this study only estimated k_{AT} of the right limb, and to corroborate the comparison between measures, k_{vert} of the right leg was also measured. Correlations between these two assessments and others are discussed below (6.3.). Despite a familiarization session on visit one, there were observed differences in the participant's abilities to perform the 5-H task. These variances in unilateral hopping coordination patterns could be a result of endurance runners not traditionally practicing this movement technique.

The observed differences in the relationships of the stiffness of each leg to that of performance, could in part be explained by the concept of leg dominance, a factor that was not assessed in this research. Previous research by Dalleau et al. [29] postulated that while running, one leg behaves like a spring (i.e. more propulsive) while the opposite leg was labelled the less propulsive or 'stick' leg. Dalleau and colleagues reported that in participants in their study, one leg performed more positive work than the contralateral limb, showing a mean asymmetry of $19.8 \pm 15.3\%$. Their observation offered a viable explanation for the mean asymmetry ($19.0 \pm 20.0\%$) between k_{vert} in the present study. Furthermore, in a sample of highly-trained sprinters (200 & 400 m specialists) who habitually make high velocity left directional turns, Churchill et al. [163] showed that the sprinters required greater peak vertical and resultant force from the right leg compared to the left. An examination of inter-limb differences however, was beyond the scope of this thesis.

Throughout the reviewed literature, different forms of hopping techniques have been instructed (maintaining straight or bent knee) to assess k_{vert} . Thus it may not be suitable to draw comparisons on stiffness values from such a range of studies. In a full 3D kinematic and kinetic analysis, Lorimer [48] reported that hopping with a straight knee showed superior comparability to running stiffness. Pilot testing for this thesis revealed that hopping with a straight knee produced more consistent contact times among participants who were unfamiliar with the task, and therefore informed the current study procedures. It is acknowledged however, that there are differences in stiffness values calculated between constrained and unconstrained hopping frequencies [46]. Hopping in time to a metronome may deviate subjects from their preferred or natural oscillation, restricting the representation of stiffness. In the current study, the five hops were performed at maximal rather than submaximal intensity to more closely replicate the

effort given in sprinting, thereby enhancing the validity of the examination between stiffness in a hopping and sprinting task.

6.4. Comparisons between classifications of stiffness variables

The secondary aim of this study was to explore the relationships between measures of stiffness in order to understand further how each measure is related in MD trained runners. The most notable finding was that there was an *extremely large* ($r = 0.92$) correlation between k_{vert} (on the right limb) and k_{leg} during overground sprinting. Practically, this finding can be interpreted that the 5-H protocol, presented for the first time in this study, may serve as an effective tool to quantify stiffness that is representative of the leg during a maximal sprinting task. Although k_{AT} had the strongest single relationship with RE in the present findings, in comparison to the other methodologies, *in vivo* measurement was by far the most time consuming and involved highly expensive equipment. As such, results of this study suggest global stiffness, as measured during maximum sprinting and unilateral hopping, appear to be useful surrogates for assessing the elastic contribution to high velocity running. The relatively simple assessments of hopping on a contact mat and the capture of contact times via high-speed video (as a comparable tool to the Optojump system [5]), can be performed in a field-based setting and undertaken with relatively inexpensive equipment, thus making it a highly feasible for practitioner use. The links between stiffness and performance reason that the global assessments are an important consideration in the mechanical profiling across a multitude of sports settings, and especially for athletes looking to improve v_{max} and RE.

6.5. Additional mechanical measures and their associations with running performance

As presented in *Chapter 2 (Table 2-1)*, the calculation of $k_{\text{leg/time}}$ required the capture of temporal gait parameters. These parameters have been reported in the literature as having contributed to the mechanical determinants of performance [42, 55]. Therefore, along with the investigation of k_{leg} , the spatiotemporal characteristics that inform its calculation were evaluated separately, and enabled an evaluation with the performance measures of RE and v_{max} .

Originally proposed by Kram and Taylor [164], metabolic energy relative to the weight of active muscle, should be inversely proportional to the time spent producing force in each step - namely, contact time (t_c). The current study results revealed both t_c and t_f had *unclear* relationships to submaximal RE and were similar to that reported when these variables informed k_{leg} . However, clear correlations with other spatiotemporal gait parameters were found, including a *large* ($r = 0.63$) association between step frequency (Sf) and RE, along with an expected negative reciprocal relationship ($r = -0.64$) between step length (S_L) and RE. A similar, yet lesser association has previously been reported in distance runners [31], where males ($N=39$) showed *small* and *moderate* correlations with RE for Sf and S_L respectively. During the maximal sprinting task in this study, t_c showed a *very large* negative correlation ($r = -0.75$) with v_{max} . Weyand et al. [99] reported limits to v_{max} in a sample of track and field athletes were not imposed by the maximum GRFs (measured on a force instrument treadmill) but through the minimum time required to apply the large, mass-specific forces necessary for propulsion. Therefore, accurate assessment of ground contact timing may predict force absorption and propulsion capacities of sprinting [165]. The use of multiple ground-based photoelectric cells (e.g. OptoJump) rather than single force plates, can provide such information in a field setting.

Anthropometric measurements that influence running efficacy, [12, 77, 136, 166, 167], and have attempted to explain the dominance of East African distance runners [77, 141] have been debated. The current study found that ankle moment arm (MA) had an *unclear* relationship with RE and v_{max} . This finding contrasts with the results from a study examining similar athletes [31], which revealed a *very large* association between ankle MA and RE. Scholz [136] stated a smaller MA at the ankle affords the AT force and stretch to be increased while running, and consequently converts a greater percentage of kinetic energy into elastic strain energy. Conversely, the work by Sano et al. [77] questioned this claim, as the MA among a cohort of Kenyans was larger compared to Japanese runners, pair-matched for body height. Therefore, this measure of MA as an indicator of performance or RE remains uncertain.

6.6. Limitations

Comparisons of the method employed in this cross-sectional study and previous studies on lower-body stiffness and performance demonstrate that several limitations exist. The study presented in this thesis attempted to follow several pre-established methodological

blueprints. However, at times situational constraints required that these methods be adapted, and these adaptations have been acknowledged through this thesis. The major limitations of this thesis are outlined below and should be taken into account when interpreting the results.

- The sample size was restricted to 11 male MD runners, due to the limited number of athletes available that met the performance criteria of “highly-trained” and “experienced”. As a result of this small sample size, several findings that related stiffness to running performance measures were *unclear*, and the uncertainty around results (indicated by confidence intervals) was often large. Nonetheless, this sample size is common when assessing specialist trained runners [29, 30].
- The results of this study are restricted to male highly-trained runners, and the results cannot be applied to a female population, runners or otherwise. Women are biomechanically different to men during running, possessing greater step rates and shorter step lengths compared with male counterparts [168]. Moreover Eiling et al. [169] reported menstrual cycle hormones (i.e. oestrogen levels) influenced MTU stiffness in netball players, resulting in fluctuations of anterior knee joint laxity across their cycle.
- While data collection purposely occurred during the competition phase of the track season, ensuring the study of athletes who were well-conditioned for track running, it was only practical to take a snapshot of the athlete’s stiffness, and this may not be representative of their stiffness values outside the track racing period.
- Given the variability of runners chosen event specialization (e.g. 800 m versus 5000 m), it was likely that the volume and intensity of running training would be different between athletes at the time of testing. However, given the small standard deviation around the mean of running distance undertaken each week (90.8 ± 15.6 km·week), it was assumed that runners at this level performed similar training loads at this time of the athletics season.
- Due to availability constraints, it was not possible to have a dynamometer familiarization session on a separate day. While dynamometer use was a relatively simple task for the participants to master, the adoption of multiple familiarization trials immediately preceding the recorded MVICs may have influenced the slack (and thus resting length) of the tendon prior to contraction. This ‘slack’ was reported to vary greatly in a group of healthy participants [170]. However, familiarization on the day of testing is consistent with previous studies [33, 34, 126].

- In the plantarflexion task, the torque produced against the dynamometer about the ankle joint was not uniquely contributed from the triceps surae and Achilles tendon. As in running and hopping, plantarflexion involves the co-contraction of the tibialis anterior, acting as the antagonist to the posterior musculature of the triceps surae. Therefore, the task undertaken was an assessment of the entire lower-limb acting to produce a plantarflexion movement.
- The use of ultrasonography during MVICs was a novel methodology in the laboratory of this institution. A substantial amount of time was spent becoming familiar with this measurement tool to ensuring the best possible effectiveness of MVIC image capture. However, the overall reliability of the expression of k_{AT} was found to be ‘moderate’ (see *Appendix 2*), based on a previously published method of determining reliability [171]. In the MVIC analysis, I did not account for tendon cross-sectional area, angle of pennation of lower-limb muscles or other 3D movements outside the sagittal plane, all of which have shown to influence stiffness [172]. Further, as the angle of the probe to the fascicles can affect 2D estimates of elongation, best attempts were made to standardize this, but small rotations of the probe may have occurred. Moreover, it was necessary to exclude one participant from the k_{AT} analysis as his large gastrocnemius muscle belly caused the probe to lose full skin contact that resulted in a distorted image.
- This study captured stiffness and metabolic data simultaneously to improve the validity of the comparisons. Nonetheless, there is the possibility that the physical constraints of running with a gas analysis mouthpiece and hose attached may cause minor kinematic alterations in running technique compared to the more familiar and unrestrained overground running.
- Due to availability limitations, the current study did not perform a 3D kinematic and kinetic analysis during submaximal or maximal running. It is difficult therefore to quantify if stiffness differences within individuals were expressed equally across the stiffness expressions of the hip, knee or ankle joints.
- Finally, this study sought to examine the influence of stiffness on RE and v_{max} . However, other factors such as mass distribution, muscle fibre type and foot strike patterns are also likely to affect these measures [12], but they were beyond the scope of this thesis.

6.7. Practical Applications and Future Directions

To improve running performance, it is worthwhile to gather information that possibly underpins mechanisms to this efficiency, for example, lower-body stiffness. Although the current small sample size was a limiting factor and uncertainty remains around relationships of global and component stiffness, there were a number of notable results. Foremost, the current findings suggest the following:

- The presented data supports previous scholarship that demonstrates a beneficial relationship between a greater lower-body stiffness and superior running economy and maximum velocity. As such, training for MD runners that targets the development of stiffness characteristics could positively affect performance.
- While stiffness of the Achilles tendon component had the strongest single relationship with running economy, global stiffness measures in maximum velocity sprinting (primarily) and maximal unilateral repeated hops (secondly) appear to be useful surrogates for Achilles stiffness that can be performed in field-based settings with relatively inexpensive equipment.

Future research could employ these more practical measures of assessing stiffness, which may assist in tracking neuromuscular changes as a mechanism underlying running performance. It may also be beneficial for future research to take multiple measures of stiffness across different phases of MD runner's training regimes. These additional measures can assess how the differences in training volume, intensity, velocity and surface affect stiffness over an extended period. Due to availability limitations, this project did not perform a 3D kinematic and kinetic analysis during submaximal or maximal running. It is therefore challenging to quantify if stiffness differences within individuals were expressed equally across the kinetic chain. Accordingly, future research should aim to further understand running performance through an in-depth assessment of the SMM incorporating k_{joint} alongside an assessment of tendon and muscular properties across the entire lower-limb. Finally, measurement of k_{AT} *in vivo* requires agreement amongst researchers so that results can be produced utilizing a common and reliable method to understand the upper and lower limits of optimal stiffness for a given task.

6.8. Conclusion

The present study demonstrated a clear association between runners who express a greater lower-body stiffness, alongside a faster and more economical running performance. This relationship was shown in agreement using both global models and structural component assessments. This study speculates that the properties of the MTU in the lower-limb, specifically Achilles tendon stiffness, explains a substantial proportion of the variance in both RE and v_{\max} in a homogenous group of highly-trained male MD runners. Though this study did not examine stiffness changes over a period of time, there is some evidence from the literature which suggests over a given period improvements of lower-body stiffness may enhance performance. As such practitioners may want to focus on strategies to improve lower-body stiffness that will result in concomitant changes in running performance outcomes.

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APPENDICES

7.1. Appendix 1: Participant Entry Criteria

Participants personal best performances from the current athletics season or season prior to the study were sourced from the Athletics New Zealand Rankings website [173] and times converted to points using the IAAF Scoring Tables of Athletics [144]. The Scoring Tables of Athletics are based on exact statistical data and conform to the following principles:

- i. The scores in the tables of different events cover equivalent performances. Therefore the tables can be used to compare results achieved in different athletics events.
- ii. The tables are progressive, which means that the same improvement of results at higher levels lead to a greater increase in scores. The degree of progressivity in the running events is different than in the jumping and throwing events due to biomechanical reasons [144].

Table 7-1. Race performance entry criteria for the present study (male).

Event	800-m	1500-m	Mile	3000-m	5000-m
750 points performance time (min:s)	2:00	4:09	4:29	8:57	15:20

7.2. Appendix 2: Reliability testing of muscle-tendon stiffness

7.2.1. Protocol

A test-retest between-day protocol on two separate days, seven days apart was performed. Participants were requested to keep any exercise in the week prior to and following the first training session the same, in order to eliminate the effects of variations in training.

7.2.2. Participants

Participants were taken from the pool of those who took part in the main study if they were able to attend a second muscle-tendon stiffness testing session. Nine out of the eleven participants from the cross sectional study were used for the test-retest assessment of reliability.

7.2.3. Method

The calculation of muscle-tendon stiffness was performed according to the methods described in full in *Chapter 4* of this thesis, specifically 4.3.2. The k_{AT} for each participant was then represented as the mean of the three MVIC trials.

7.2.4 Statistical Analysis

Descriptive statistics including group means and standard deviations were calculated for stiffness and for moment and elongation variables separately. Using the methods of Hopkins [174], data were assessed for between trial measurement reliability and measurement variability at the 90% confidence level following log transformation to allow results to be expressed as percentage [104]. Robustness was maintained by using two criteria each to determine the level of reliability and variability [103].

Average reliability was determined to be ‘good’ if the percent difference between means (MDiff%) was <5% and the effect size (ES) was *trivial* (0 – 0.2) or *small* (0.2 - 0.6) [102]. If one of these criteria were not met then measurement reliability was interpreted as ‘average’. ‘Poor’ reliability resulted in neither criteria being met [103]. Measurement variability was assessed from typical error, reported as coefficient of variation percentage (CV%) [103, 104] and intra-class correlation coefficient (ICC) with upper and lower confidence limits [103]. Criteria for ‘small’ measurement variability were CV <10% [103] and ICC >0.70 [103, 105]. If CV was >10% or ICC <0.70 then variability in the

measurement was deemed ‘moderate’. ‘Large’ measurement variability was reported if neither criteria for ‘small’ was met.

For overall reliability of our measure of k_{AT} (effect size, percent difference between means, coefficient of variation and interclass correlation) were assessed. The assertion of ‘Good’ reliability required all four criteria to be met. ‘Moderate’ reliability resulted from one criteria outside the limits, while if two or more criteria were outside the limits, a ‘poor’ overall reliability was recorded [45].

7.2.5. Results

The reliability and variability statistics are reported in *Table 7-2*. The measure of k_{AT} , when expressed as the mean stiffness of the three MVIC trials, was used in this thesis. This expression had good average reliability (MDiff% <5% and ES < 0.6) and indicated overall reliability to be moderate. This expression was compared to k_{AT} defined as the linear slope of the force–elongation curve created from all individual trials of the MVICs and submaximal ankle PFs. Although this later stiffness estimation appears more commonly in the literature [17], in our laboratory, reliability of this method was poor. It was thus preferred to estimate k_{AT} from the mean peak stiffness attained across three MVICs.

Table 7-2. Summary of the reliability for various expressions of muscle-tendon unit stiffness in male middle distance runners (N=9).

<i>k_{AT}</i> expression	Pearson's Correlation (CL)	Reliability (MDiff%)	Effect Size	Variability CV% (CL)	ICC (CL)	Overall Reliability
<i>Mean</i> of 3 x MVICs	0.85 (0.54 - 0.96)	1.8 (-12.4 – 18.3)	0.05 (-47.1 – 47.2)	18.7 (13.1 - 34.1)	0.87 (0.61 – 0.96)	Moderate
<i>SLOPE:</i> All trials (25-100% MVIC)	0.71 (0.21 – 0.91)	15.4 (-5.3 – 40.6)	-0.35 (-48.3 – 47.6)	25.3 (17.6 – 47.1)	0.74 (0.32-0.92)	Poor

CL = confidence limits at the 90% level, MVIC = maximal voluntary isometric contraction, SLOPE = the slope of the linear regression of force and MTU elongation

7.3. Appendix 3: Ethical Approval



21 October 2014

Andrew Kilding
Faculty of Health and Environmental Sciences

Dear Andrew

Re Ethics Application: **14/324 Measurements of lower-limb stiffness and the relationships to economy and velocity in well-trained middle-distance runners.**

Thank you for providing evidence as requested, which satisfies the points raised by the Auckland University of Technology Ethics Committee (AUTECS).

Your ethics application has been approved for three years until 21 October 2017.

As part of the ethics approval process, you are required to submit the following to AUTECS:

- A brief annual progress report using form EA2, which is available online through <http://www.aut.ac.nz/researchethics>. When necessary this form may also be used to request an extension of the approval at least one month prior to its expiry on 21 October 2017;
- A brief report on the status of the project using form EA3, which is available online through <http://www.aut.ac.nz/researchethics>. This report is to be submitted either when the approval expires on 21 October 2017 or on completion of the project.

It is a condition of approval that AUTECS is notified of any adverse events or if the research does not commence. AUTECS approval needs to be sought for any alteration to the research, including any alteration of or addition to any documents that are provided to participants. You are responsible for ensuring that research undertaken under this approval occurs within the parameters outlined in the approved application.

AUTECS grants ethical approval only. If you require management approval from an institution or organisation for your research, then you will need to obtain this.

To enable us to provide you with efficient service, please use the application number and study title in all correspondence with us. If you have any enquiries about this application, or anything else, please do contact us at ethics@aut.ac.nz.

All the very best with your research,

A handwritten signature in black ink, appearing to read 'K O'Connor', is written over a light grey background.

Kate O'Connor
Executive Secretary
Auckland University of Technology Ethics Committee

Cc: Simon Rodgers simon.rogers@aut.ac.nz; Chris Whatman; Simon Pearson

A u c k l a n d U n i v e r s i t y o f T e c h n o l o g y E t h i c s C o m m i t t e e

WA505F Level 5 WA Building City Campus

Private Bag 92006 Auckland 1142 Ph: +64-9-921-9999 ext 8316 email ethics@aut.ac.nz

7.4. Appendix 4: Participant Information Sheet

Participant Information Sheet



Date Information Sheet Produced:

23 September 2014

Project Title

Measurements of lower-limb stiffness and the relationships to economy and velocity in well-trained middle-distance runners.

An Invitation

My name is Simon Rogers and I am a Masters student at AUT University, based at AUT-Millennium. On behalf of my supervisors Associate Professor Andrew Kilding, Senior Lecturer Dr Chris Whatman, and biomechanist Dr Simon Pearson, I would like to invite you to assist us in our research on middle-distance running. We are conducting research to determine the reliability of measurements of lower-limb stiffness and the relationships to running economy and speed. This will allow us to gain a greater insight into improving performance and monitoring risks of injury with greater relevance to high performance middle-distance runners.

It is entirely your choice as to whether you participate in the project or not. If at any time you decide you no longer want to participate, you are free to withdraw from the study up until the completion of data collection without consequences. Your consent to participate in this research will be indicated by your signing and dating the consent form. Signing the consent form indicates that you have read and understood this information sheet, freely given your consent to participate, and that there has been no coercion or inducement to participate by the researchers from AUT.

What is the purpose of this research?

This research will investigate biomechanical stiffness of the lower-body, which has been previously linked to running performance and injury. Lower-body stiffness is reliant on the properties of multiple joints and their associated tendons during running. This project aims to enhance our understanding of how these relate to key measures relevant to middle-distance running performance. This will allow practitioners to track an athlete's stiffness and better understand the links between specific techniques and indicators of running performance and risk factors of injury. Specifically, this project will investigate how stiffness relates to running economy and maximum sprint speed in trained middle-distance runners. These results will also produce a Master thesis with the view of it being published in a scientific journal.

How was I identified and why am I being invited to participate in this research?

You have been invited to participate because you are a well-trained middle-distance runner, who competes over distances from 800-5000 metres and are aged between 16-35. You have a race performance time (obtained from Athletics NZ Membership Manager, Brett Addison) which scores at least 750 points on the IAAF scoring table. These are the first two criteria to be considered for participation. If you also have no current injuries, or have not been injured within the past three months, you are eligible to participate.

What will happen in this research?

You will be required to visit the SPRINZ lab on 3 occasions. During the first session, your height and weight and some leg length measures will be taken. You will then have a number of reflective balls attached to your skin (where possible) or clothing using non-toxic tape and elastic bandages. You will be able to wear your own light-weight running shoes which will also have reflective dots attached. You will carry out a five minute warm-up on a stationary bike and then you will be asked to perform a number of lower limb tasks that will be recorded by an ankle plate, 2D camera's will capture your running technique.

Specifically, during visit one, you will perform a strength task which involves lying flat on a table, where you will have your right foot secured in an ankle bracket and be asked to push down with the ball of your foot into the fixed bracket as hard as you can for 3 seconds. During these ankle movements, we will take an ultrasound image of your calf which involves applying some gel and positioning a surface probe on your Achilles tendon. Following this, you will be asked to perform 5 consecutive vertical jumps on a locked treadmill which is inlaid into the laboratory floor. The final part is running on this treadmill at 4 submaximal speeds for 4 minutes at each speed.

In the second session on the following day, you will repeat the 5 jump test, then perform 3 maximal 50-m sprints on an indoor track. Finally you will perform a maximal oxygen consumption test on a treadmill to measure your running. The test will start easy and progressively get harder, pushing you to exhaustion. You will be given time to become accustomed to these tasks before we record your movements.

As part of testing the repeatability and reliability of our measures, we will ask some participant's to return for a third session, 7 days after the first visit, to repeat tests (ankles strength task, 5 vertical jumps, submaximal treadmill running for at 4 speeds for 4 minutes at each speed). Each session will be no more than 1 hour.

What are the discomforts and risks?

There may be some discomfort (VO_{2max}) and risk associated with measures in this research, but this will not be beyond your normal high intensity level of running. You will be asked to complete the tasks in normal light weight running clothing. All tasks have been selected to ensure that they do not exceed your ability. If you are experiencing discomfort at any stage you are encouraged to inform the researcher with you at the time in order that they can best address the problem. If you have any questions regarding and risk or comfort that you anticipate, please feel free to address these concerns to the researcher so that you feel comfortable at all times throughout the process.

What are the benefits?

As a token of our appreciation for participating in the research you will be given the opportunity to have your movements assessed for the purpose of performance enhancement. You can also have a free report of oxygen consumption test to aid in your training goals. We will make a contribution towards the costs of your travel, consisting of a \$15 petrol voucher

What compensation is available for injury or negligence?

In the unlikely event of a physical injury as a result of your participation in this study, rehabilitation and compensation for injury by accident may be available from the Accident Compensation Corporation, providing the incident details satisfy the requirements of the law and the Corporation's regulations.

How will my privacy be protected?

All your personal details will remain confidential and will only be available to Simon Rogers and the research supervisors during the period of the study. Your face be included in the 2D video footage, so you will not be identifiable from the video data. All data will be stored on password protected computers or in locked files. Following completion of data analysis, your data will be stored by the SPRINZ research officer in the secure SPRINZ ethics storage. It is possible that all data collected may be used for other purposes in the future, especially as it is not uncommon for analysis techniques to advance or change over the years, therefore data will be stored indefinitely for this purpose.

What are the costs of participating in this research?

You will be required to attend the SPRINZ laboratories at AUT-Millennium on two consecutive days, for one hour on each occasion. In addition, for further reliably testing, 12 randomly selected male participants will also be asked to return for a third one hour session 7 days following their first test day to repeat the stiffness measures. All participants will only perform one maximum oxygen consumption test (on the second lab visit).

What opportunity do I have to consider this invitation?

We would appreciate it if you could let us know within one week whether you would be available to take part in the study or not. After consideration you may withdraw your participation at any time.

How do I agree to participate in this research?

If you agree to participate please fill in the attached consent form and return to Simon Rogers.

Will I receive feedback on the results of this research?

If you would like feedback on the results of the research please indicate on the consent form.

What do I do if I have concerns about this research?

Any concerns regarding the nature of this project should be notified in the first instance to the Project Supervisor, Associate Professor Andrew Kilding, andrew.kilding@aut.ac.nz

Concerns regarding the conduct of the research should be notified to the Executive Secretary of AUTEK, Kate O'Connor, ethics@aut.ac.nz 921 9999 ext 6038.

Whom do I contact for further information about this research?

Researcher Contact Details:

Simon Rogers, Sport Performance Research Institute New Zealand, School of Sport and Recreation, AUT University, Private Bag 92006, Auckland 1020, Mobile: 027 6323 138, simon.rogers@aut.ac.nz

**Approved by the Auckland University of Technology Ethics Committee on 21 October 2014,
AUTEK Reference number 14/324.**

7.5. *Appendix 5: Consent form*

<h1>Consent Form</h1>	 <p>AUT UNIVERSITY <small>TE WĀNANGA ĀRDNUI O TAMAKI MĀKAU RAU</small></p>
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*Project title: **Measurements of lower-limb stiffness and the relationships to economy and velocity in well-trained middle-distance runners***

*Project Supervisor: **Associate Professor Andrew Kilding***

*Researcher: **Simon Rogers***

- I have read and understood the information provided about this research project in the Information Sheet dated 23 September 2014.
- I have had an opportunity to ask questions and to have them answered.
- I am not suffering from heart disease, high blood pressure, any respiratory condition (mild asthma excluded), any illness or injury that impairs my physical performance.
- I understand that I may withdraw myself or any information that I have provided for this project at any time prior to completion of data collection, without being disadvantaged in any way.
- I understand there may be some discomfort associated with measures in this research, but this will not be beyond your normal high intensity level of training.
- I am not suffering from any illness or injury that may prevent me from being able to complete the tasks detailed in the information sheet.
- I agree to provide and allow my movements to be recorded using 2D video analysis.
- I agree to allow the researcher to view information from my physiotherapist/sport medicine provider regarding my musculoskeletal injury history.
- I agree to take part in this research.
- I wish to receive a copy of the report from the research (please tick one): Yes No

Participant's signature:

Participant's name:

Participant's Contact Details (if appropriate):

.....
.....
.....
.....

Date:

**Approved by the Auckland University of Technology Ethics Committee on 21 October, 2014
AUTEC Reference number 14/324**

Note: The Participant should retain a copy of this form.

7.6 Appendix 6: Participant Advertisement



SPORTS PERFORMANCE
RESEARCH IN NEW ZEALAND AT AUT MILLENNIUM INSTITUTE



Invitation Trained Athletes

My name is Simon Rogers and I am a Masters student at AUT University.

Under the guidance of my supervisory team, Associate Professor Andrew Kilding, Dr Chris Whatman, and Dr Simon Pearson we will be conducting research to determine the reliability of different measurements of lower-limb stiffness and their relationships to running economy and running speed. This research will allow us to gain a greater insight into improving performance and monitoring risks of injury with greater relevance to high performance middle-distance runners (800m – 5000m).

Eligibility Criteria:

If you meet all the following criteria, you are eligible for this study.

- Aged 16 and 35 years old,
- Currently train **at least** 5 days per week,
- Do not currently have a lower-extremity injury **and** have not had an injury in the past three months,
- Race over 800-5000-m, and your best time from the last **two years** meets the standard in **one or more** events listed below

	800-m	1500-m	Mile	3000-m	5000-m
Men	2:00.8	4:09.2	4:28.8	8:56.6	15:20.4
Women	2:25.2	5:03.4	5:26.3	10:56.5	18:56.5

What the research will involve:

2 (for Women) and 3 (for Men) visits to the SPRINZ Laboratory at **AUT Millennium** for:

- Lower limb strength and power tests
- Video Analysis of your running technique
- Speed test on the indoor track
- VO_{2max} and running economy test

Time Required:

2 – 3 hrs in total.

Benefits to the Athlete:

You will be given results and feedback on all assessments performed including your VO_{2max}. These might aid your training/performance. You will also be provided with a \$15 reimbursement for your travel costs to the lab.

Would you like to participate?

If you would like to find out more information or register your interest to take part in this study, please contact me for a detailed participant information sheet and consent form.

Thank you for your consideration

Simon Rogers BSci, PGDip.Sp&Ex, Masters Candidate Sport Science
Sports Performance Research Institute New Zealand | AUT University
AUT-Millennium, 17 Antares Place, Mairangi Bay, 0632, NZ
Email: simon.rogers@aut.ac.nz | Mob: 027 632 3138